

DOCTORAL THESIS

Dynamic properties of the lumbar spine in people with non-specific low back pain

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Dynamic properties of the lumbar spine in people with non-specific low back pain

by

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**A thesis submitted in partial fulfilment of the requirements for
the degree of PhD**

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Abstract

Non-specific low back pain (LBP) has been associated with alterations in the biomechanical properties and muscle activities of the trunk, but it is unclear how these changes are related to the dynamic stability of the trunk. During sitting, the structures of the trunk stabilise the upper body counterbalancing external moments acting on the trunk.

The aim of this research was to evaluate a range of biomechanical variables including the hip and lumbar spine joints range of motions, moments and powers the viscoelastic properties of the trunk, and the role of the muscles while a participant was performing a balancing task in sitting and to compare results between healthy and LBP subjects.

A custom-made swinging chair was used to perform the balancing task. It was designed to challenge primarily the trunk and to minimise the effect of the lower limbs so that the role of the trunk could be examined in isolation.

Twenty-four participants with LBP and thirty healthy participants were requested to sit on the custom-made swinging chair and to regain the balance after tilting the chair backward for 10° and 20°. Electromagnetic motion track system sensors were placed on the participants' back, one at the sacrum level and one at the first lumbar vertebral level to measure hip and lumbar kinematics. One further sensor was placed on the chair to track its rotation, which was also the rotation of the lower limbs. Forces data were taken from a force-plate which was mounted at the bottom of the chair.

Inverse dynamic equations were used to derive the muscle moment acting at the hip and lumbar spine joints using data from the force platform and the motion tracking system. Muscle power was then calculated by multiplying the muscle moment and the corresponding joint angular velocity. Trunk viscoelastic parameters were derived using a second order linear model combine trunk moment and motion.

Chair motion and trial duration were used to evaluate dynamic stability and task performance, in particular, the angular displacement of the chair was fitted in an equation describing the underdamped second-order response to a step input to derive natural frequency and damping ratio and to evaluate possible differences between groups.

Activities, reaction times and co-contraction of the trunk muscles were evaluated using surface electromyography (EMG). The surface electrodes were placed bilaterally on the erector spinae, rectus abdominus, external and internal oblique.

Kinematic analysis showed that the hip range of motion increased whereas spine range of motion angle decreased in participants with LBP for both tilt angles ($p < 0.05$). No significant differences were found in muscle moment and power between healthy and LBP subjects ($p > 0.05$). The duration of contraction of various trunk muscles and co-contraction were significantly longer in the LBP subjects ($p < 0.05$) when compared to healthy subjects, and the reaction times of the muscles were also significantly reduced in LBP subjects ($p < 0.05$).

Trunk stiffness was found increased for LBP subjects ($p < .05$) while no difference was found for damping coefficient. There were no significant differences between the 2 subject groups in the time required to regain balance, and in the dynamic stability parameters, the natural frequency and damping ratio.

The present study showed LBP was associated with alterations in biomechanical variables; in particular stiffness, hip and lumbar spine joints kinematic and muscle responses were altered in subjects with LBP when compared with healthy group. However, these alterations did not affect dynamic stability and moment developed at joints level, suggesting that LBP subjects adopted a different strategy to maintain balance but with the same effectiveness as the healthy subjects without any worsening of the symptoms. This may suggest to clinicians to encourage patients to remain active

rather than to avoid movements. On the other hand, compensatory strategies were achieved with increased co-contraction at the expenses of muscle efficiency. This may lead to muscle fatigue and increase in spinal stress. Future research should clarify if the observed biomechanical alterations in this study are consequences or causes of LBP; or if the biomechanical changes and pain operate in a vicious circle, reinforcing each other leading to chronic conditions. This would help achieve our ultimate goal of developing effective treatment strategies, and it is hoped that the work of this thesis has helped us take a significant forward towards this goal.

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Publications

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To my grandmother Vera.

1 INTRODUCTION

1.1 Problem statement

Non-specific low back pain (LBP) is one of the most widespread pathological conditions affecting humans; it is believed that nearly everyone has experienced this condition at least once in their lifetime (Savigny et al., 2009). Every year in the UK, it is estimated that at least 30% of the population has an episode of LBP and 20% of these cases will consult their GPs (Macfarlane et al., 2006).

The chronic nature of the problem is such that 60% of the population with LBP continue to suffer one year on from the first episode. It has been also estimated that 16% of the LBP population were not able to work because of the pain, and of these, 30% were still unable to work after 6 months (Hestbaek et al., 2003).

LBP has been defined as pain localised in the lower part of the spine, between the 12th thoracic vertebrae and the sacrum (Krismer and van Tulder, 2007) and people with LBP can be divided into three categories, also known as “diagnostic triage”: specific spinal (e.g. trauma, degenerative condition as osteoarthritis), nerve root pain/radicular pain, and non-specific LBP if symptoms are caused by an unknown condition (Waddell, 1987). Another important classification for LBP is related to symptom duration: in accordance with the literature LBP has been defined as “acute” if symptoms lasted less than 6 weeks, “sub-acute” if it lasted between 6 weeks and 12 weeks and “chronic” if it lasted more than 3 months (Krismer and van Tulder, 2007, Savigny et al., 2009).

It is estimated that only 15% of LBP patients have a specific diagnosis for the pain (e.g. fractures, structural deformity, radicular pain). As a result, there is a lack in treatment rationalisation in the majority of non-specific LBP and this may contribute to a decreased quality of life and an increased cost for the community (Airaksinen et al., 2006). The latest estimation of the cost of back pain in UK was completed in 1998 and it

indicated that the healthcare cost for back pain treatment was £1632 million of which 35% was the cost of non-NHS healthcare (Maniadakis and Gray, 2000, Savigny et al., 2009). In addition, a further estimated £9090 million were lost due to production loss i.e. inability to work, which further magnifies the significant economical impact of LBP (Maniadakis and Gray, 2000).

LBP subjects exhibited alterations in different biomechanical variables as it has been showed by several studies. Lumbar spine motion was found decreased (Shum et al., 2005b, Shum et al., 2007a) and trunk moment distribution altered while performing everyday life activities (Shum et al., 2007b); reduced postural control was found in LBP subjects when compared with healthy subjects while performing balancing task (Cholewicki et al., 2000a, Preuss et al., 2005, Radebold et al., 2001); increased trunk muscular activities and co-activations were found in LBP subjects during different activities (e.g. lifting exertions, trunk flexion) (Granata and Marras, 1995b, Marras et al., 2001, Marras et al., 2004, Radebold et al., 2000).

Furthermore, mechanical characteristics of the trunk, such as viscoelastic properties, showed to be affected by LBP. Intrinsic properties of the spinal structure in both in-vitro (Panjabi, 1992) and in-vivo studies (Colloca and Keller, 2001, Latimer et al., 1996c) showed reduction in spinal stiffness and damping coefficients.

1.2 Thesis aims

The aim of this study was to evaluate possible alterations in biomechanical variables and how these alterations, if present, may be related to changes in mechanical properties of the trunk and, in turn, affect dynamic stability.

Kinematic and kinetic variables of the lumbar spine, trunk muscular activities were measured during unstable sitting, while a second order linear model was employed to

derive viscoelastic parameters of the trunk and dynamic stability assessed by modelling trunk response to the proposed activity.

Data were then compared between LBP and healthy subjects to examine possible differences between the two groups.

During the balancing task it was expected that subjects would alternatively flex and extend the trunk to find the balanced position; it was hypothesised that alterations would be present in lumbar spine kinematic, kinetic variables and in trunk muscles activity as a result of LBP, and these alterations would affect trunk viscoelastic properties. On the contrary, it was hypothesised that dynamic stability would not be affected by LBP and, as consequence, this would imply that LBP subjects would use a different strategy to perform balancing task, but this strategy would be as effective as the strategy used by healthy subjects.

Results of this study will increase knowledge about LBP and related biomechanical mechanisms in people with non-specific pain. This will help clinician in improving rehabilitation program and treatment for back pain, and in turn, increase outcomes for the patients, decrease symptoms related to LBP, and restoring trunk biomechanical function.

1.3 Thesis outline and objectives

In chapter 1 an introduction, purposes and outline of the thesis are presented.

In chapter 2 a review of the literature it is presented. The purpose of this chapter is to review methodologies used for lumbar spine kinematic and kinetic investigation, measuring of trunk muscular activity, viscoelastic modelling of the trunk, thoracolumbar curvature measuring and assessment of dynamic stability. The review explains the

choice of the technique for this study. Furthermore this chapter provides a review of existing literature regarding biomechanics of the trunk in healthy and LBP subjects. Limitations in the current methodology are evaluated and the design of the current study has been made to address these limitations and provide more useful data to clinician.

In chapter 3 a description of the design and building of the custom made swinging chair used for this research will be presented. Furthermore the inverse dynamic equations used for the trunk kinematic and kinetic analysis and inserted in a custom made MATLAB® (R2007b, MathWorks Inc.) will be presented.

In chapter 4 methodology and results about biomechanical response of the trunk is presented, whereby trunk displacement and external moment acting on the trunk are used to develop a second order linear model to derive the viscoelastic parameters of the trunk while a subject is performing a balancing task. The viscoelastic properties of the trunk would be examined in a sitting position in this experimental study, while the participants will try to regain a balanced position after been tilted on a swinging chair. This approach challenges the trunk function without the influence of the lower limbs and enables us to quantify possible differences between healthy and LBP participants in regard to their viscoelastic properties.

In chapter 5 methodology and results about dynamic stability analysis of the trunk during unstable sitting in healthy and LBP participants are presented and discussed. The purpose of this study is to evaluate the dynamic stability and kinematics of the lumbar spine during unstable sitting, to model the response using a second order

equation and to determine the differences in the model parameters between healthy and LBP subjects.

This would allow us to calculate the damping and natural frequency characteristics of the response, Furthermore subjects' performance will be evaluated in terms of their ability to regain balance using the duration and accuracy of regaining balance. Kinematics of the spine and the hip are evaluated to understand their relative contributions to balance control. This experimental approach could provide new insights into the biomechanics of the spine in unstable sitting and to re-examine some of the discrepancies in the findings of previous work.

In chapter 6 methodology and results about analysing trunk kinetic and trunk muscular activity in LBP and healthy subjects are presented and discussed. The purpose of this study is to examine the muscular activities and kinetics of the trunk during unstable sitting and to determine differences in these responses between healthy and LBP subjects.

The nature of the muscle contraction employed to counterbalance the external moment has not been established. Inverse dynamic analysis has been used to determine if the trunk muscles are generating a flexor or extensor moment to counterbalance the external moment, and the analysis of muscle powers permitted us to evaluate whether the muscular contraction is eccentric or concentric. However, the inverse dynamic model does not allow us to study co-contraction of muscles which may not present during the task

This limitation is addressed by collecting EMG data from trunk muscles, to evaluate any possible co-contraction and to better understand how various muscles contribute to the muscle moment and power. The methodology explained in the chapter it has been proposed to show the association between LBP and changes in muscle responses.

In chapter 7 a general discussion about the findings is presented. The purpose of this chapter is to summarise thesis results and to discuss results from the three different aspects of the study together in order to enhance knowledge about LBP and to give direction for future research.

In chapter 8 conclusions are presented. The purpose of this chapter is to explain the contribution to the knowledge about LBP that this thesis gave and how results presented may benefit LBP sufferers.

1.4 Scope and boundaries

It is within the scope of this thesis to evaluate the trunk biomechanics in a sitting position, while the participants are trying to regain a balanced position after been tilted on a custom made swinging chair and to compare data between healthy and no-specific LBP subjects.

This study offers the first step towards the development of a methodology to be used as a biofeedback training exercise for people with non-specific LBP for restoring the motor control by providing evidence of the usefulness of the use of the custom made swinging chair to challenge trunk stability and to evaluate progress of the rehabilitation training.

The experimental setup is design in order to simulate common activities such as sitting on a bus or in a car. This allows us to challenge mainly the trunk, to minimise the influence of other body parts (e.g. lower limbs).

It is not in the scope of this study to evaluate the effect of any clinical interventions on non-specific low back pain subjects. Instead, trunk mechanical properties, muscular behaviour and dynamic stability are evaluated during the proposed task and compared

to find possible differences between non-specific low back pain subject and control. To achieve this scope, different biomechanical variables are investigated during the proposed task as follows:

- Motion of the lumbar spine and hip are evaluated using electromagnetic motion tracking system to detect possible differences in trunk motion strategies
- Kinetics of the hip and lumbar spine are evaluated to investigate possible alteration in joint loading sharing strategies and joint powers developed;
- Electrical activity of trunk muscles is collected using surface EMG to evaluate muscular activity, latency and co-contractions;
- Stiffness and damping coefficients are derived combining trunk moment and angular displacement data in a second order linear model to evaluate possible differences in trunk viscoelastic properties;
- Subject's performance in the proposed task is investigated in order to evaluate how possible alterations in biomechanical variables due to non-specific low back pain may impact their ability to regain balance on a custom made swinging chair after external perturbation.

Human spine physiology and anatomy are beyond the scope of this thesis and the reader is expected to have basic understanding of trunk biomechanics and non-specific low back.

As already stated above, it is in the scope of the study compare biomechanical variables between subjects with non-specific low back pain and healthy subjects, as a results participants are recruited and inclusion and exclusion criteria are established to exclude subjects with any confirmed pathological spine condition, different than non-specific low back pain.

2 LITERATURE REVIEW

2.1 Introduction

The purpose of this review is to give a context to this research, to illustrate current knowledge and gaps in understanding the relationship between trunk biomechanics and non-specific LBP and to explain how this research fills these gaps, increasing understanding of LBP and related biomechanical mechanisms.

In the first section of this review the anatomy of the trunk is briefly presented, in particular a description of the bones, muscles and ligaments composing the spinal structure are described.

In the following part of this review, different aspects of the trunk biomechanics are explored, such as:

- Trunk kinematics
- Postural control
- Dynamic stability and muscle functioning
- Trunk viscoelastic properties
- Trunk muscular activity
- Trunk kinetics
- Thoracolumbar curvature

Current methodology used to assess a certain aspect in in-vivo studies, together with current knowledge of the effects of non-specific LBP are presented and revised for each aspect listed above.

In the last section a review of current clinical management option is presented, in order to show the current understanding of the action mechanism that explains how a certain therapy improves condition for non-specific LBP.

2.2 Anatomy of the trunk

2.2.1 Spinal structure

The spinal column runs from the base of the skull to the pelvis and it is formed by a series of individual bony structures called vertebrae that support the weight of the body and protect the spinal cord.

The spine is divided into regions: cervical, thoracic, lumbar, sacral and coccygeal. The vertebrae forming the cervical, thoracic and lumbar regions are movable while vertebrae forming sacral and coccygeal regions are fused.

Between each movable vertebra there is an intervertebral disc that is formed by an outer fibrous ring (annulus fibrosus) and an inner soft and elastic material (nucleus pulposus). This disc absorbs vertical stress and permits the movement of the vertebral column. Each vertebra is formed by different parts:

- The body: the weight bearing part, shaped as a disk, with a rough inferior and superior surface where the intervertebral discs lie;
- Vertebral arch: the posterior part of the vertebra and it forms, together with the body, the vertebral foramen; the vertebral foramen of all vertebrae form the spinal canal that protects spinal cord; in the vertebral arch there are also two short thick processes, the pedicles that project posteriorly from the body to fuse with the laminae;
- Processes: there are seven processes that arise from the vertebral arch; two transverse processes (extending laterally on each side) and one spinous process (extending posteriorly) are used for muscle insertion. The two superior articular processes articulate with the two inferior articular processes of the adjacent vertebra, forming the facet or intervertebral joints.

Ligaments help to hold vertebra together, stabilising the spine; in particular anterior and posterior longitudinal ligaments are continuous bands that run for all the length of the spinal column along the vertebrae bodies and prevent excessive motion. Another ligament is the ligamentum flavum that connects the lamina of the vertebra protecting the spinal cord and avoiding excessive vertebral movements.

The spinal cord runs through the spinal canal; it is part of the central nervous system and its primary function is to connect the brain with the rest of the body, but it can also independently control reflexes and central pattern generators (Tortora and Derrickson, 2008).

2.2.2 Abdomen and lumbar muscles

Posture, motion and other trunk actions are performed with the independent or coordinate activity of various muscles. The abdominal wall is formed by four pairs of muscles: rectus abdominis are superficial muscles that are involved in flexion of the vertebral column and in compressing the abdomen. External obliques are superficial muscles present in the left and right side of the abdomen region which act bilaterally in flexion of the vertebral column and in compressing the abdomen and singularly for lateral flexion and contralateral rotation of the vertebral column. Internal obliques are intermediate muscles present in the left and right side of the abdomen region which act bilaterally in flexion of the vertebral column and in compressing the abdomen and singularly for lateral flexion and to rotate the vertebral column to the same side. Transversus abdominis are deep muscles that compress the abdomen. Another muscle that is present bilaterally in the lumbar region is the quadratus lumborum; this is a deep muscle that helps extension of the lumbar portion of the vertebral column when both sides act together and singularly acts in lateral flexion of the column.

On the posterior side of the lumbar region the major muscle group is the erector spinae, which forms a prominent bulge on both sides of the vertebral column. The major action of this superficial group of muscles is the extension of the vertebral column, but is also involved in controlling flexion, lateral flexion, and rotation to maintain the lumbar curvature. Another important group of muscles is the multifidus, which is a deep group of muscles involved in extension, lateral flexion and rotation of the spine and in stabilising and assisting the column during motion and posture maintenance. Small muscles are present in each vertebra: rotatores are involved in rotating the vertebral column, interspinales and intrasversarii are important for stabilisation of the spine (Tortora and Derrickson, 2008).

2.3 Trunk kinematics

2.3.1 Kinematics measurement techniques for the trunk motion

This section provides a review of the different methods to assess static and dynamic lumbo-sacral kinematics. Principle of functioning, advantages and disadvantages of each method will be presented along with results of reliability and validation test. A comparison between the methods will be carried out and the choice of the technique for this study will be explained.

2.3.1.1 Inertial sensors

Gyroscopes and accelerometers have been classified as inertial sensors but with the progress of technology the term inertial sensor generally indicates a unit formed of gyroscopes, accelerometers and magnetometers used in combination in order to decrease the limitation of each tracking system, also known as an inertial measurement unit. An accelerometer consists of a mass-spring-damper system in a rigid frame where the mass displacement due to acceleration and relative to the frame is detected in three motion axes. Accelerometers have been used to measure spinal postural change in sitting by Wong and Wong (2008b) and the reliability was evaluated comparing the data from the accelerometer with a rotational alignment device, finding that the RMS was less than 1° with a high correlation coefficient (0.99). Further studies used this technology to measure and evaluate spinal motion (Lou et al., 2002).

Gyroscope detects angular velocity measuring the Coriolis forces present in a vibrating device. The output voltage is proportional to the angular velocity and so the angular orientation can be obtained through integration (Tong and Granat, 1999).

Lumbar motion and posture has been assessed using gyroscopes, in particular a study from Lee et al. (2003), and showed high reliability (0.97-0.99) in all planes of motion. Recently inertial measurement unit has been applied to biomechanical research. Gyroscopes in the unit give information about angular orientation through integration of the angular velocities. However, this integration is subjected to drift when the sensor is stationary. Different strategies have been used to overcome this limitation (Lee et al., 2003, Mayagoitia et al., 2002) through different filtering, but with the development of the inertial measurement unit, additional information provided by magnetometers and accelerometers have been used to compensate for the error and this technique has been used for spinal motion measurement (Lee et al., 2003, Wong and Wong, 2008b) and for running and cycling measurements (Tan et al., 2008). Saber-Sheikh et al. (2010) demonstrated high similarity between electromagnetic and inertial sensor systems to measure spinal motion (0.05° different range for artificial hinge joint rotation and $0.28-0.69^\circ$ for random six degrees freedom rotation of wooden jig) while Wong and Wong (2008a) had similar results when compared with a video-optical system (maximum error for spinal motion was 3.1°) for trunk measurements. The integration drift is the main limitation of these sensors because although it can be compensated with data from magnetometers and accelerometers, it is also affected by time and temperature. Moreover data from the inertial sensors cannot be referred to an external coordinate system (Wong et al., 2007).

2.3.1.2 Video-optical based technique

Reflective markers are placed on the body of the subject and infrared digital cameras are used to record the motion of the markers. Motion analysis is done using a reference position; moreover joint moment can be measured when used in combination with a force platform. Manufacturers of such systems report very high accuracy in all three

planes of motion, but Windolf et al. (2008) showed that the accuracy is task, environment, and set-up dependent and the error can reach a maximum of 0.42 mm during a systematic test. Accuracy have been also tested comparing data from radiographs, where results showed low error from the optical system in all three plane of motion (maximum error $\pm 2^\circ$, and RMS $<1^\circ$) (Pearcy et al., 1987).

Video optical systems (VOS) have been used to evaluate trunk kinematics during gait analysis (Crosbie et al., 1997), performing different tasks such as lateral and forward bending from standing (Al-Eisa et al., 2006, Esola et al., 1996), sitting (Al-Eisa et al., 2006, Van Daele et al., 2009) and to evaluate balancing strategies in the sitting position (Van Daele et al., 2009).

This system has been also compared with other motion analysis systems and the results discussed above showed similarity about accuracy and repeatability with the electromagnetic system (Hassan et al., 2007) and inertial sensors (Wong and Wong, 2008a). VOS can give a lot of information about the motion because several markers can be added and detailed modelling of the trunk can be built. On the other hand adding markers increases the complexity of processing data and in turn, the time taken for analysis. Another limitation is related to the way as the motion is detected: each marker needs to be in line of sight of at least three cameras, so they need to be placed around the experimental environment, limiting the portability of the system. In addition, depending on the tasks, the number of the cameras needs to be increased in order to detect all the markers, which increases the cost of the system.

2.3.1.3 Radiographic motion analysis

Kinematic of the trunk has been studied in the past using planar radiographic images (Dvorák et al., 1991). The analysis was restricted to one plane of motion and the motion

was measured using a standing reference image whereby a vertebra was traced and then compared with other images taken from different positions.

This technique gave information about just one plane of motion at the time and coupled movements could not be seen. Pearcy et al. (1985) used a combination of two radiographs in order to get three dimensional information of spinal ROM, but only static measures were assessed. Cineradiography (Harada et al., 2000) and videofluoroscopy (Adams et al., 1988) have been used to measure dynamic trunk movement. The main issue with radiographic methods is exposure to dangerous radiation, especially in multiple planar radiographic and cineradiography whereby the participant is subjected to relatively high exposure to radiation. Videofluoroscopy uses less radiation but the data processing is more time consuming and the overall quality of the images is reduced. Moreover the technique is expensive and need to be restricted to a laboratory environment to decrease the risks correlated with radiation exposure.

2.3.1.4 Clinical-based techniques

One of the first, easiest and inexpensive methods to measure spinal motion is skin distraction. The positions of bony landmarks are marked on the overlying skin and the distance between two points is measured. Measurements are repeated in different positions in order to evaluate ROM of the joints. High correlation (>0.9) coefficient was found when compared with planar radiographies (Macrae and Wright, 1969) but limitations have been reported concerning influence of the height and body weight. In addition, data gives only information about motion in the sagittal plane and it cannot be used for dynamic experiments (Pearcy and Hindle, 1989). Another device cheap and easily operated is the inclinometer which gives the angle of inclination relative to a reference. In trunk studies the device was able to differentiate between pelvic and lumbar contribution to the motion (Ng et al., 2001). However, the device does not give

information about axial rotation and dynamic measures are not possible. Lumbar motion has also been assessed using the distance between finger tips to the floor but even though reliability was found to be high in a study evaluating this method (Frost et al., 1982), there were concerns that the contribution to the measured trunk motion from the pelvis, thorax and upper limbs could significantly alter the results (Wong and Lee, 2004). Another device used for assessing lumbar motion is the electrogoniometer, which is made with two ends plate connected by a strain gauge wire or a potentiometer which are sensitive to motion variation. The electrogoniometer has been validated comparing with lateral radiographic images for standing measures (Perriman et al., 2010), and in spinal motion (Bible et al., 2010). This method also allows data collection from dynamic experiments, but it can only measure two planes of motion and there are constraints with suitable attachment sites for the two plates onto the body. Moreover the system gives data of the relative motion between the two ends of the plate without the possibility of referring to an external reference.

2.3.1.5 Electromagnetic motion tracking system

The electromagnetic motion tracking system is easily operated, portable and useful tool to measure the kinematic of the lumbar spine in three dimensions. This system is formed mainly by two parts, the transmitter and the receiver both wired to a system electronic unit. An electromagnetic field is created by the transmitter, which is also used as reference for the position of the receiver. The receiver detects the field generated by the transmitter and in response generates a signal that is input into a mathematical algorithm to compute the receiver position. It is possible to detect the position of more than one receiver at the same time. The static accuracy of this technology has been reported in term of RMS as 0.8 mm for position and 0.15 ° orientations. The system has been found to be useful in kinematic studies (An et al., 1988). In particular Mills et al.

(2007) evaluated the repeatability of the dynamic motion measurement performed with an electromagnetic system during walking, showing high intra-trial, intra-day/inter-tester, inter-day/intra-tester reliability. High repeatability was also found for spinal motion measurement during fast bowlers in cricket (Burnett et al., 1998). Total RMS error was found less than 0.2° by Pearcy and Hindle (1989) for dynamic measures of spinal rotation. Electromagnetic devices have also been compared with other technology as radiographic imaging and video-optical motion system. Hassan et al. (2007) used a mechanic arm in order to evaluate reliability of optical and electromagnetic motion system. Regarding the magnetic system, the differences between measured motion and the actual motion of the arm was found to be -1.23° , -0.95° and 0.37° in the three planes of motion. Differences between electromagnetic and video-optical system was 0.15° , -0.32° and 0.54° for the three planes of motion, showing high similarity between the two techniques.

Electromagnetic motion system has also been compared in an in-vitro experiment with lateral radiography for intervertebral motion and the results showed small differences between the two methodologies in the sagittal plane rotation ($0.47 \pm 0.24^{\circ}$) (Zhao et al., 2005). In addition, Yang et al. (2005) found reliability in measuring spinal angle with electromagnetic system when compared with radiographic methodology.

The error due to relative motion between the sensors placed on the back skin and the vertebrae has been evaluated and it was found less than 8% of the total motion when compared with radiographic images (Yang et al., 2008). Also Bull and McGregor (2000) evaluated the error due to relative motion between the skin (where the sensor is placed) and the vertebrae in spinal flexion/extension using MRI imaging and they found a measuring average error of $\pm 1^{\circ}$.

Lumbar spine movements have been assessed placing the sensors at level of the first lumbar vertebrae and first sacral vertebrae (Shum et al., 2005a, Shum et al., 2005b,

Shum et al., 2007a, Van Herp et al., 2000, Wong and Lee, 2004) This allowed understanding the kinematics of the lumbosacral tract. Increasing the number of sensors could give extra information about the behaviour of the lumbar spine but it will imply a increasing difficulty in the analysis. Limited operational range is one of the biggest limitations of the system. Although the manufacture report indicated 3,05 metres as the maximum distance between the receiver and transmitter, some studies revealed that this operating zone was much smaller, in particular Milne et al. (1996) results showed a cubic area around 220-720 mm³ from the source. This parameter is dependent on the manufacture, but validation tests were recommended in order to check the operation zone. Metal interference is another limitation of this technique. The presence of metal in the experimental environment could affect the measure and reduce the range of functioning (Milne et al., 1996). Comparison with radiographic methodology showed that electromagnetic system was reliable and accurate for biomechanical studies and it carried less risk for the subjects. Furthermore the system is cheaper than other technologies such as the video-optical motion capture system. The limited range of functioning and the sensitivity to metal interference are the main limitations of the system.

2.3.1.6 Comparison between methods

Clinical based methods are quick and inexpensive way to evaluate ROM of joint and basic lumbar kinematics. Radiographic has been used to evaluate the lumbar motion but it appeared to be more useful for static measures and it carries significant health risks to the participants. Inertial sensors, electromagnetic system and video-optical based systems now are becoming more popular in the field of biomechanics when evaluating kinematics of the human body. All three methodologies have been shown to provide reliable and accurate measurements of trunk kinematics. However all these

methodologies have different constraints that make them useful for different purposes: video-optical based method can only be used in a laboratory environment and is more expensive than the other two. The information given can be more detailed by adding more markers but this will cause an increase in cost and analysis time. Inertial sensors are subjected to limitations of the device i.e. integration drift and it is difficult to use in combination with other instruments e.g. force platform. Electromagnetic system is relatively cheap and easily operated to handle, it can be synchronized with other devices (e.g. force platform and EMG), and it is not dangerous for the patients. Limited operation range and incompatibility with metallic environments/objects are the limitations of this system

In conclusion an inertial sensor would be more useful to evaluate kinematics in everyday situations outside the laboratory environment. A video-optical method is more useful for research that can be performed in a laboratory environment such as gait analysis. Electromagnetic system can be used in both laboratory and no-laboratory environment and allows integration with other devices but it has a limited range of operation and can be affected by metal. For these reasons it is very useful for clinical examinations and for experiments that involve limited motion e.g. sitting or posture tests and when other parameter such as muscular activity need to be evaluated.

In this research the electromagnetic motion system was chosen for measuring the kinematics of the lumbar spine, in particular the 3SPACE FASTRAK® from Polhemus Inc. was used. The range of operation needed for this study was relatively small, compatible with the operational features of the FASTRAK®, because the task involved participants sitting on a custom made swinging chair, trying to balance on it. No compatibility constraints were present because the swinging chair was built using wood that avoid any interference with the electromagnetic field and presence of metal implants was an exclusion criterion for participants. Before a final decision on the

method to use in this research, validation tests were performed and they showed the compatibility of the experimental environment with the tracking system (Chapter 3).

FASTRAK® was also chosen because it allowed synchronisation with other devices such as EMG system and force platform that were needed for this study. Validation tests (Chapter 3) were performed to assess the synchronization between the devices.

It can be argued that also the video optical measurement system, in particular VICON system that was available in this laboratory, offered the same advantages as for the electromagnetic system. The FASTRAK® system was preferred because firstly it did not need all the calibration and the post analysis that is required for the VICON system; secondly because the markers used in the optical video system and applied on the participant needed to be seen at least from three cameras at the same time while performing the experiment and the design of the swinging chair made this difficult for markers be placed on the lower part of the back i.e. risk of losing fundamental information during the data collection was high and changing the chair design or the setup of the VICON system was less convenient than using the FASTRAK® system.

2.3.2 Kinematics evaluation for trunk motion

Kinematics evaluation can include joint ROM, velocity, coordination, motion pattern which can vary related to the tasks performed by the subjects. Normal kinematics of the trunk has been evaluated with participants performing simple trunk movements as flexion, extension and rotation and ROM of the lumbar and spinal joints have been used to assess mobility in the subjects.

Early studies evaluated the 3-dimensional intervertebral ROM using static radiographic images, taken with the participants in different positions (Pearcy et al., 1987). Marras and Wongsam (1986) considered the trunk as a whole segment and evaluate the ROM

and velocity while subjects were performing trunk flexion, extension at normal and maximum velocity.

In the recent studies kinematics has been evaluated in dynamic tasks using methods described in the previous section (Esola et al., 1996, Mayer et al., 1984, Paquet et al., 1994, Wong and Lee, 2004). Dynamic ROM of the hip and lumbar spine joints were evaluated in different studies where ROM and contribution to the motion were evaluated while participants were performing different tasks and in different posture, e.g. standing or sitting; in particular Wong and Lee (2004) evaluated the 3 dimensional ROM of the hip and lumbar joints and joints contribution to the motion using lumbar spine/ hip motion ratio in different trunk tasks in standing position; results showed that ROM in forward bending was higher for hip joint indicating higher contribution to the motion from the hip movements for forward bending; similar results were found for twisting while opposite results were found for side bending with lumbar ROM higher for lumbar spine and. ROM for the two joints was similar for backward bending, but the ratio was 1.36, indicating a lumbar spine higher contribution to the movements. Also Esola, McClure et al. (1996) investigate ROM and contribution of the hip and lumbar spine joints to the motion: participants bended forward until they reached the maximum angle while the motion was recorded; ROM results were in agreement with Wong and Lee (2004), with lumbar spine motion smaller than hip motion. Lumbar to hip flexion ratio was calculated for different bending forward angle intervals (0-30°, 30-60°, 60-90°) and the data showed that lumbar spine had a greater contribution on the first part (0-30°) of the motion, while after the hips became predominant. Al-Eisa et al. (2006) evaluated the ROM and the symmetry of the trunk performing a lateral flexion and axial rotation using an optoelectric motion analysis system with sensors on the vertebrae T1, T6, L1,L5, and sacrum and compared the results between these two tasks in standing and sitting. Results showed that the ROM for both thorax (defined as the segment from T6 to L1)

and lumbar spine (defined as the segment from L1 to L5) in standing is higher than in sitting while performing lateral flexion but it is lower than in sitting while performing axial rotation. Another study involving sitting was performed by Van Daele et al. (2009), where a participant was sitting on the flat side of a wobble board with knee and hips flexed at 90 degrees. Pelvic segment motion (defined by the midpoint between the markers on the PSISs and the 2 markers on the left and right iliac crest) and trunk segment motion (defined by the midpoint between the markers on the left and right iliac crest and the 2 markers on the left and right acromion) were evaluated. Results showed that total angular deviation in the three planes of motion calculated for pelvic and trunk segments were in the same range (sagittal plane 64° for trunk and 72° for pelvis; transverse plane 60° for trunk and 61° for pelvis; frontal plane 43° for trunk and 42° for pelvis).

It can be concluded that ROM and the contribution to the motion of the trunk joints are task and initial position (e.g. sitting or standing) depend, as the results showed differences in these variables related to this parameters.

Another variable evaluated in kinematics studies of the trunk is joints velocity; in the studies of Wong and Lee (2004) and Esola, McClure et al. (1996) described above, showing differences in this parameter related to the task.

Coordination between hip and lumbar spine joints has been investigated through cross correlation between hip and lumbar spine motion patterns to assess the similarity in the motion pattern and time lag to evaluate phase difference between the two movements. . In Wong and Lee (2004) study, correlation coefficients were above 0.84 and time lag negligible for all the tasks indicating high degree of association and no phase shift between joints motion. Similar results were found for Van Daele et al. (2009) in sitting, where correlation between trunk and pelvis pattern were evaluated and results showed high coordination (Pearson correlation coefficient > 0.9) between the movements for all

the planes of motion. The trunk movements explored in the above studies are important to evaluate trunk mobility and to investigate the effects of several conditions to those variables. However trunk kinematic demonstrated to be task dependent (Larivière et al., 2000a), therefore it is also important to investigate trunk motion strategies while participants perform other tasks and in particular everyday activity in order to measure the motion and the contribution of each trunk part (e.g. hip and lumbar spine joints) and to evaluate possible alteration due to a certain condition. A common activity that has been used in trunk kinematics research is putting a sock from a sitting position. Shum, Crosbie et al. (2005b) placed electromagnetic sensors on the thighs, L1 and S1 vertebrae and recorded the motion while participants were putting a sock sitting on a chair. ROM, velocity and ratio between lumbar spine and hip motion to evaluate joint contribution to the motion were measured for all the planes of motion. Results showed that lumbar motion was smaller than hip motion (e.g. mean of the lumbar spine angle 48° , while for hip angle 114° in sagittal planes) for all the planes of motion, which was also confirmed by the ratio that was found <1 indicating a higher contribution of the hip to the motion. Also velocity was found higher in the hip joint for all the planes of motion. Other activities have been used to evaluate as picking-up an object from the floor during sitting from contralateral and ipsilateral side (Shum et al., 2007a), sit-to-stand and stand-to-sit (Shum et al., 2005a). In both the studies the lumbar spine motion was found smaller than hip motion in all plane of motion (e.g. in sagittal plane for ipsilateral picking the lumbar spine ROM was 48° and 93° for the hip). Also lumbar spine/hip motion ratio was <1 in all the activities showing a higher contribute of the hip to these activities (Shum et al., 2005a, Shum et al., 2005b).

Walking is a common and important task that has been also used to evaluate impairment level than could be associated with a certain condition (Fairbank and Pynsent, 2000, Savigny et al., 2009). It is important to investigate the kinematic of the

trunk during walking in order to understand the contribution of the upper body to this task and to evaluate possible relation with decreasing of performance and spinal condition.

Kinematics of the upper body has been investigated during walking in a laboratory condition using a treadmill and video optical motion track system (Crosbie et al., 1997, Lamothe et al., 2002, Selles et al., 2001). Lamothe, Onno et al. (2006) evaluated the rotational amplitudes of the motion for pelvis, lumbar and thorax segments and the coordination between the segments using the continuous relative Fourier phase between lumbar and pelvis motion and thorax and pelvis motion. Participants performed several trials at different speeds in order to evaluate the effect of velocity on the variables. Results showed, in agreement with other similar studies (Crosbie et al., 1997, Lamothe et al., 2002), that the motion in the sagittal planes was almost absent for all the segments as expected; transverse and frontal planes motion values (e.g. for frontal planes: thoracic 7.2°, lumbar 6.4° and pelvis 6.3° at comfortable speed) were much smaller than values from the other studies reviewed above because the trunk was not the main structure involved in this task. Amplitude of pelvis rotation increased in both transverse and frontal planes with velocity.

2.3.3 Kinematics and LBP

LBP has been associated with reduced spinal range of motion that can lead to functional limitation and disability, and decreasing quality of life (Savigny et al., 2009). Everyday life activity such as putting a sock can be more difficult for people with LBP because of the reduced dynamic mobility of the trunk (Strand and Wie, 1999). Reduced ROM of the trunk has been found in LBP subjects while performing trunk bending (Marras and Wongsam, 1986) forward and cycles of flexion/extension movements (Pearcy et al., 1985). Other studies evaluated the differences between healthy and LBP

subjects in the hip and lumbar spine joints ROM while performing different trunk movements (Al-Eisa et al., 2006, Esola et al., 1996, Mayer et al., 1984, Paquet et al., 1994, Wong and Lee, 2004). In particular Wong and Lee (2004) investigated the differences between healthy subjects and two LBP groups, subjects with and without limitation in straight leg raise (SLR), finding that motion of the lumbar spine in the sagittal plane was decreased significantly for forward bending from 61° to 33° for LBP and to 30° for LBP with SLR. Similar results for the lumbar ROM were found for lateral bending and axial twisting in their principal planes of motion (frontal plane for lateral bending and transverse plane for axial twisting). Forward bending showed also a significant decrease in the hip ROM. Hip ROM for LBP subjects with SLR was also significantly decreased for backward bending. Different results were found by Esola, McClure et al. (1996) for forward bending: no significant differences in ROM for hip and lumbar spine joints were found between healthy and LBP groups; the contrasting results may be explained by differences in the LBP group characteristics: while Wong and Lee (2004) evaluated people with current LBP, Esola, McClure et al. (1996) recruited people with history of LBP in the previous 5 years but without any symptoms for at least 2 weeks. This difference may imply that the alteration of ROM of the joints could be a strategy to reduce pain and risks of further damage while a patient is experiencing the pain (Wong and Lee 2004).

Results of the ROM of the lumbar spine and hip joints reported above are consistent with results found in studies evaluating the ROM of these joints during everyday activities; decreased in ROM for lumbar spine joint was found in LBP participants while they were performing tasks as picking-up an object from the floor during sitting from contralateral and ipsilateral side (Shum et al., 2007a), sit-to-stand and stand-to-sit (Shum et al., 2005a) and putting a sock from a sitting position (Shum et al., 2005b). Furthermore during picking-up from sitting position and for sit-to-stand and stand-to-sit

activities also the hip joint ROM was found decreased (Shum et al., 2005a, Shum et al., 2007a).

Van Daele, Hagman et al. (2009), in the unstable sitting experiment described above, found that ROM of the pelvis and trunk segments were increased for LBP subjects that was in contrast what has been described so far. In the experiment also postural sway and trunk/pelvis motion correlation were evaluated and they were both increased in LBP subjects. Increased active stiffness of the trunk muscles in order to prevent pain and unwanted movements were used to explain differences in correlation and postural sway. Controlling of the unstable seat was more difficult with increased trunk stiffness (Van Daele et al., 2009). In walking no significant differences were found for hip and lumbar spine joints ROM between LBP and healthy subjects (Lamoth et al., 2006, Lamoth et al., 2002, Selles et al., 2001). Joints velocity has been also evaluated in kinematics studies that compared LBP and healthy participants. Trunk velocity was decreased in forward bending in LBP participants (Marras and Wongsam, 1986). Hip and lumbar spine velocities were decreased in LBP subjects for bending (Paquet et al., 1994) but also in everyday activity as described above (Shum et al., 2005a, Shum et al., 2005b, Shum et al., 2007a). As for the ROM parameters, joint velocities were similar between symptomatic and asymptomatic subjects by Esola, McClure et al. (1996), but different subjects characteristics (history of LBP instead of current LBP) may explain these results.

No significant differences were found in joint velocity during walking between symptomatic and asymptomatic participants (Lamoth et al., 2006).

The contribution of the lumbar spine motion to the overall trunk motion was found significantly reduced in LBP patients with and without limitation in SLR for side bending and axial rotation, while for forward bending the reduction founded in the two LBP groups was not significant (Wong and Lee 2004). Esola, McClure et al. (1996) in the

study presented above, investigated the spine/hip ratio during forward bending from 0° to 90° dividing the tasks in three sub-phases of 30° degrees and they found a significant decreased contribution of the lumbar spine motion in the 30° to 60° sub-phase for people with history of LBP. Differences in the contribution were also evaluated for everyday activity and significant decreasing in lumbar spine contribution was found in LBP subjects with and without limitation in SLR during sit-to-stand, stand-to-sit and while putting a sock from a sitting position (Shum et al., 2005b). No differences were found while subjects performing picking-up activity (Shum et al., 2007a).

Decreased coordination between trunk structures has been found in LBP subjects; Wong and Lee (2004) showed r decreasing in the correlation strength between hip and lumbar spine joints, along with differences in motion phase in LBP participants, especially for trunk twisting (correlation decreased from 0.87 to 0.71 in LBP subjects). Cross-correlation was also used in the studies from Shum, Crosbie et al. (Shum, Crosbie et al. 2005a; Shum, Crosbie et al. 2007a) for evaluating coordination of hip and lumbar spine motions (e.g. lumbar flexion with hip flexion, lumbar rotation with hip abduction, etc.) in putting a sock and picking-up activities and alterations in the LBP subjects, especially in the group with SLR limitation, were founded. As mentioned earlier in this section, the correlation between trunk and pelvic segments motion was increased for chronic LBP patients and it was explained as a strategy to increase spinal stiffness (Van Daele et al., 2009). Similar results were found in walking where thoracic-pelvic and lumbar-pelvic coordination was found increased in chronic LBP subjects (Lamoth, Meijer et al. 2006).

Asymmetry between left and right side was evaluated in standing and sitting by Al-Eisa, Egan et al. (2006) using sensors placed on the centre, left and right side of T1, T6, L1, L5, and S1 vertebrae. In detail, asymmetry was evaluated for thorax and lumbar spine as the maximum differences in length between the left and right side of these

segments, which was calculated with the data from the sensors placed on the sides of the vertebra. Thoracic asymmetry was found similar for both group of subjects while lumbar asymmetry was found increased in symptomatic subjects. Other variables related with tasks, as speed or time to complete the tasks were also used. LBP exhibited differences in the walking speed: for Selles, Wagenaar et al. (2001) the comfortable walking speed was reduced in LBP participants (average for LBP group was 2.9 km/h while for healthy group was 3.8 km/h); in the study of Lamothe, Meijer et al. (2006) while all healthy participants were able to walk at high speed (e.g. 6.2 and 7 km/h), just 68% for 6.2 km/h and 26% for 7 km/h of LBP participants were able to walk at those high speeds.

Time to complete the task for basic trunk motions such as bending and twisting and for everyday activity as putting a sock, sit-to-stand, stand-to-sit and picking-up objects was found significantly increased for LBP participants (Shum, Crosbie et al. 2005a; Shum, Crosbie et al. 2005b; Shum, Crosbie et al. 2007a; Wong and Lee 2004), for example in the putting a sock tasks the mean of the time required to complete the test was 4.8 s for healthy group while for LBP was 8.0 s (Shum, Crosbie et al. 2005b).

2.4 Postural control

According with (Horak, 1987, Mergner and Rosemeier, 1998), postural control can be explained as *“the ability to maintain equilibrium in a gravitational field by keeping or returning the centre of body mass over its base of support”* in a particular conditions (e.g. standing or sitting) (Horak, 1987, Mergner and Rosemeier, 1998).

Different systems such as visual, vestibular and somatosensory system, contributes to postural control, coordinating the motor response of the human body to maintain the equilibrium (Mergner and Rosemeier, 1998, Radebold et al., 2001). using inputs from the eyes and from the vestibular organs in the ears are useful for postural control along with , feedback from different receptors in the different body tissues, which, for postural control, evaluated position and motion of the musculoskeletal system (Mergner and Rosemeier, 1998).

Postural control has been evaluated in order to investigate how different conditions (e.g. Parkinson's disease, LBP, Multiple Sclerosis) may lead to impairment in this system (Lanzetta et al., 2004, Radebold et al., 2001, van der Burg et al., 2006).

2.4.1 Postural control evaluation

Postural control has been evaluated in upright standing in healthy and chronic LBP participants. Mientjes and Frank (1999) evaluated the root mean square (RMS) of the centre of pressure (COP) trajectory for the anterior-posterior and medial-lateral directions while the participants were asked to lean forward as far as they could without losing balance. The test was repeated in different condition (eyes closed/open; feet on a stable or on a foam surface; head normal or tilted backward) in order to evaluate control, challenging the three systems that are involved in the postural control. Results showed that in both anterior-posterior and medial-lateral directions the RMS increased while participants were standing on foam comparing with the test performed on a stable

surface; no significant differences were found for eyes closed and head tilted conditions. The increased RMS of the COP variables suggested that postural control during leaning forward from standing position was more affected when the proprioceptive input was challenged than when visual and vestibular systems were manipulated (Mientjes and Frank, 1999). Similar results were found by Della Volpe, Popa et al. (2006), that tested postural control with participants standing on a movable platform with eyes open and closed. The test was done with the platform parallel to the floor and tilted and results showed increased COP velocity and RMS in the anterior-posterior direction when the base was tilted; more dramatic increasing in these variables were found when closed eyes and tilted base conditions were applied together.

Percentage of participants that successfully performed a balancing task was used by Mok, Brauer et al. (2004) to evaluate postural control in different conditions: participants were asked to stand over a force platform and to hold bilateral stance for 70 seconds and unilateral stance for 30 seconds with eyes closed and opened. Results showed that all participants were able to perform the bilateral task, with eyes open or closed while for the unilateral task the percentage decreased to 45% when the subjects had eyes closed. Differences were also found in the COP motion and velocity between open and closed eyes condition: the mean of the COP range of motion was increased from 22mm to 55mm and mean of COP velocity was increased from 3.2mm/sec to 10.2 mm/sec, indicating how the visual impairment could affect postural control.

Postural control has been also evaluated while where participants were performing tasks in a sitting position that involved the trunk motion rather than the lower limbs motion. In this manner trunk postural control could be isolated and evaluated independently and alterations due to different condition that may not be seen in standing experiment may be present in sitting experiment (Cholewicki et al., 2000a).

Radelbold, Cholewicki et al. (2001) used chair with foot support to prevent lower body movements; moreover the base of the chair was designed to allow for attachment of hemispheres with different diameters to have four level of instability. The experiments was performed with eyes open and closed to evaluate the dependence on visual feedback. Participants were asked to maintain equilibrium for 7 seconds. Results showed that for the highest level of instability with the eyes open all the participants were able to finish the task, while with the eyes closed the 29% of the participants could not maintain the equilibrium for 7 seconds. RMS and peak of the anterior-posterior and medial-lateral COP was found increased with seat instability.

Other variables were used to investigate trunk postural control, as endurance time, percentage of successful trials, COP range of motion and trunk kinematics and used to evaluate alteration due to a certain pathological condition e.g. Parkinson's disease (van der Burg et al., 2006).

2.4.2 Postural control and LBP

Postural control experiments were used to evaluate possible alterations due to LBP. LBP patients demonstrated increased postural sway (measured as range of motion of the COP) while performing postural control experiment in standing position (Hamaoui et al., 2004, Mientjes and Frank, 1999).

Della Volpe, Popa et al. (2006), in the experiment described earlier in this section, found that the anterior-posterior COP RMS and velocity were different between healthy and LBP patients when participants were asked to perform the experiment with the support surface rotated; in particular LBP subjects demonstrated to increase anterior-posterior sway to maintain postural stability when visual and vestibular information were challenged (eyes closed and base rotated) and they were relying more to proprioceptive

system, which may imply an alteration of the proprioceptive feedback from the lumbar spine due to LBP.

LBP patients demonstrated also poorer balancing in unilateral stance experiment: Mok, Brauer et al. (2004) evaluated unilateral and bilateral stance with eyes closed and open showing that LBP participants achieved a reduced number of successful balance tasks. Furthermore in the study researchers evaluated the hip strategy (detected as the anterior-posterior shear force) and the ankle strategy (detected as the COP displacement) contributions to the balancing tasks. Results showed that the hip strategy contribution was decreased in LBP participants, suggesting that the decreased performance in LBP subjects may be due to this alteration that, in turn, may be related with altered muscle control and proprioceptive impairments of the lumbar spine. Ruhe et al. (2012) found a relation between pain relief and decreasing in postural sway in standing position in LBP subjects.

Possible alterations due to LBP in postural control has been evaluated also in sitting position because, as mentioned above, sitting position permitted to isolate the trunk from the lower limbs and to evaluate the effect of the pain in the lumbar spine (Cholewicki et al., 2000a).

Radebold, Cholewicki et al. (2001), in the study mentioned above, evaluated trunk postural control in unstable sitting and trunk muscles response in a sudden load test in order to evaluate possible alteration in the motor control due to chronic LBP. Total COP path length/s was used as postural control index. In the sudden loading test participants were sitting in a semi-seated position and load were attached to the trunk and released without warning while EMG was recording trunk muscular activity that was used to derive average of onset time of muscles. Results about the trunk postural test showed that at the most difficult level, the percentage of LBP participants that were able to finish the test was 69% for eyes open, against 100% for healthy participants and 13% with

eyes closed, against 71% for healthy participants indicating poorer balancing and higher dependence to visual feedback for LBP subjects. Furthermore COP motion was found increased in LBP and this alteration increased more with difficulty and removing visual feedback, indicating that vestibular and proprioceptive feedbacks were less effective in trunk postural control for LBP patients than in healthy people. The sudden loading test showed that onset time response of the trunk muscles was significantly increased in LBP subjects. Onset time was also correlated with the postural control index and significant positive correlation was for the postural test with eyes closed that may indicate that alteration in motor control in trunk muscle may contribute to impairments in trunk postural control. These results were in accordance with what was found by Van Daele, Hagman et al. (2009) in a similar experiment about unstable sitting. Results showed increased postural sway for chronic LBP subjects when compared with healthy subjects. Furthermore pelvis and trunk motion were evaluated and the results showed increased motion for both segments and also increased correlation between motions of the two segments in symptomatic participants, implying that pelvis-trunk stiffness was higher because the two segments were moving together almost as one segment.

Van Dieen, Koppes et al. (2010) found no differences in postural sway while participants were performing a sitting balance experiment, that was in contrast with findings from Radebold, Cholewicki et al. (2001); this may be explained by the condition of the LBP participants: van Dieen, Koppes et al. (2010) recruited people with self reported current and recent LBP while Radebold, Cholewicki et al. (2001) were evaluating people with chronic LBP (at least 6 months symptoms duration); this may also indicate that postural control impairments may be related with aggravation of the LBP condition.

2.5 Dynamic stability and muscles functioning

2.5.1 Dynamic stability of the spine

The spinal column is not the only structure involved in the stability of the spine. Panjabi (1992) found three essential subsystems that are involved in the process: the spinal column, spinal muscles and tendons and the neural control unit (Figure 2-1).

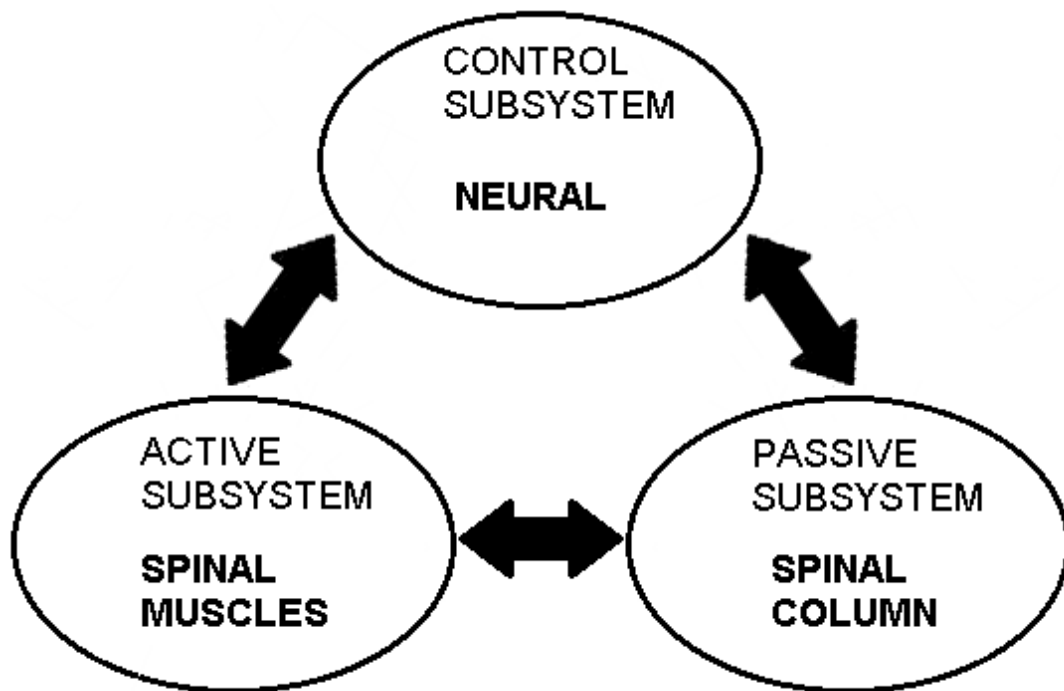


Figure 2-1: Spinal stability subsystems (Panjabi 1992).

The spinal column consists of the intervertebral discs, bone, and the ligaments and it provides mechanical support to the trunk. However this passive subsystem itself is not enough to provide spinal stability, in fact in-vitro experiments found that an intact lumbar specimen buckles when it is subjected to an average load of just 88N (Crisco et al., 1992). Moreover it was found that this structure had a greater role in stability when the

spine reaches the ends of its ROM as the ligaments increase their resistance to movement.

A second active subsystem, formed by trunk muscles and tendons, generates forces in order to provide stability to the spine. Tendons give a feedback to the neural system regarding magnitude of force generated by the muscle. The effective stiffness of the trunk is the sum of the effects of the active and passive system, controlled by the neural control unit. Studies showed that the major contribution to spine stability is given by the stiffness of the trunk muscle that is proportional to muscle activation (Moorhouse and Granata, 2005). Evidence showed that stability was increased with increasing muscular activity, which is achieved through co-contraction of agonist and antagonist muscles during active exertions but with disadvantages of increasing load in tissue and energy consumption (Cholewicki et al., 1997, Gardner-Morse et al., 1995, Gardner-Morse and Stokes, 2001, Esola et al., 1996).

Different studies have established that the muscular component provides a significant contribution to trunk stiffness (Bergmark, 1989, Cholewicki and McGill, 1996, Cholewicki et al., 1997, Gardner-Morse and Stokes, 1998, Panjabi, 1992). Furthermore these studies have shown that only a relatively small amount of muscular activation can give a great contribution to trunk stiffness and stability. A recent study also established that no single muscle has a greater contribution to stability over any others, and that the contribution is dependent on the loading magnitude and direction (Cholewicki and VanVliet IV, 2002).

The neural control unit evaluates proprioceptive feedback signals from biological transducers i.e. collecting data regarding spinal position and motion, and then coordinates muscle activation in order to stabilise or maintain stability of the spine through reflex and voluntary responses.

2.5.2 Dysfunction in the stability system

Decreased spinal stability can be due to a dysfunction in one of the spinal stabilising subsystems (Panjabi, 2003). In-vitro studies have been performed in order to evaluate different mechanical properties of the injured disks and significant changes were found after comparing data from a spinal unit before and after an induced injury (Panjabi et al., 1984). Significant differences were also found for ligaments (Panjabi et al., 1994) and facet joints (Abumi et al., 1990) with regards of mechanical properties, spinal stability and ROM due to injuries.

The active subsystem, which is formed by the tendons and muscles, may develop dysfunction as consequence of injuries or specific condition (e.g. LBP) that may influence its ability to produce and coordinate muscle tension, receive information from the neural system and to send proprioceptive feedback (Panjabi, 1992). When the passive subsystem decreases its function the active subsystem compensates by increasing its contribution, e.g. increasing muscular co-activation (Radebold et al., 2000); if also the active subsystem is compromised, it cannot compensate for the passive subsystem and the spine becomes unstable (Panjabi, 1992).

Dysfunction can also develop in the neural subsystem; the function of this system is to monitor and adjust forces of the trunk muscles, in order to adapt the activity to a particular task. Faulty feedback from the proprioceptors of the lumbar spine or faulty in the control unit may lead to excessive or inadequate muscle force, delays in response time to a perturbation that may decrease spine stability and increase risks of injuries in the spinal structures (Panjabi, 1992). Furthermore excessive force production in the muscles due to error in the neural system may lead to damage in the active system and to increased loading on the spine structures (Marras et al., 2001), which can cause damage (Adams et al., 1996). Neuromuscular control dysfunctions have been already related to patients with lumbar dysfunctions, there is some evidence that showed that

increasing body sway may be related to a reduction in efficacy of neuromuscular control system in people with low back pain or spinal canal stenosis, and as such can compromise spinal stability (Hamaoui et al., 2004, Mientjes and Frank, 1999, Panjabi, 2003).

2.6 Trunk viscoelastic properties

2.6.1 Viscoelastic properties measurements

2.6.1.1 Definition of viscoelasticity

Definitions of elastic and viscous material are needed in order to introduce the concept of viscoelasticity. An elastic material responds to a stress perturbation with a sudden deformation; the stress is directly proportional to deformation multiplied by a constant that characterises the material. This behaviour is described by the Hooke's law and it can be represented by a mechanical component called a spring (figure 1a) characterised by a stiffness coefficient K .

In a viscous material, stress is proportional to the velocity of deformation multiplied by a constant that characterises the material and the behaviour is described by Newton's viscosity relationship. It can be represented by the mechanical component called a dashpot (Figure 2-2) characterised by a damping coefficient B .

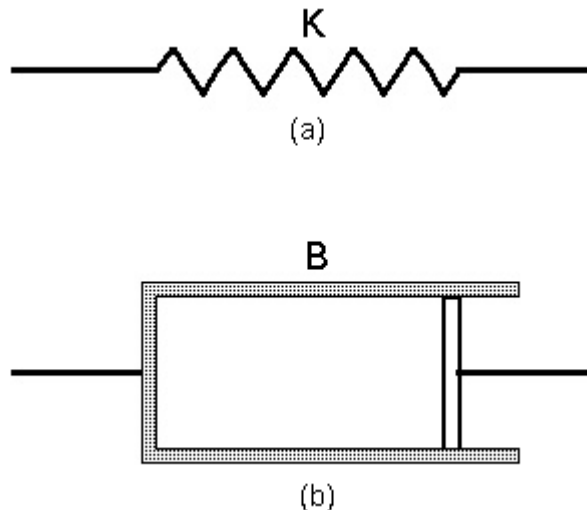


Figure 2-2 : the two mechanical components: (a) spring with stiffness coefficient K and (b) dashpot with damping coefficient B .

Most of all material behaviour is a mix between viscous and elastic response and biological tissues such as muscles, tendons, bones or blood vessels have demonstrated viscoelastic features.

Investigating the viscoelastic properties has been found useful to evaluate possible alterations in the body structures as results of certain conditions, e.g. joint injury (Panjabi et al., 1984); effects of exercising (McNair and Stanley, 1996).

Using the technique that has been well established in mechanical engineering, in some basic studies the stiffness of the trunk was evaluated as the ratio between the trunk motion and the forced applied: Scholten and Veldhuizen (1986) derived stiffness of the trunk with the participants lying down with the lower part of the body on a table and the shoulder and head on a board that was free to move; an external load was applied and the motion recorded and used to calculate the stiffness coefficient.

Hooke principle has been used to evaluate stiffness in different spinal landmarks (e.g. L3, S1 vertebrae) using a device which delivered a posteroanterior force to the lumbar vertebrae and measured the load and displacement of the skin surface where it was applied (Colloca and Keller, 2001, Latimer et al., 1996c). Accuracy of the device was tested using elastic beams with known stiffness and the difference between known value and value measured was less than 2.5%. Reliability was assessed with intraclass correlation coefficient that was reported to be 0.9 for this type of apparatus (Latimer et al., 1996a). More detailed viscoelastic models are built in order to evaluate viscoelastic properties of a certain material during a certain task. These mechanical models used the strain/stress relationship in order to derive damping and stiffness coefficients, which quantify the viscoelastic properties of the material (Flügge, 1975).

In this section different models used to evaluate viscoelastic parameters in the human musculoskeletal system will be described.

2.6.1.2 Three-parameter model

The three-parameter model has been used to evaluate viscoelastic behaviour of different materials such as polymer (Ferry, 1980) but also for biological tissues (Fung, 1993).

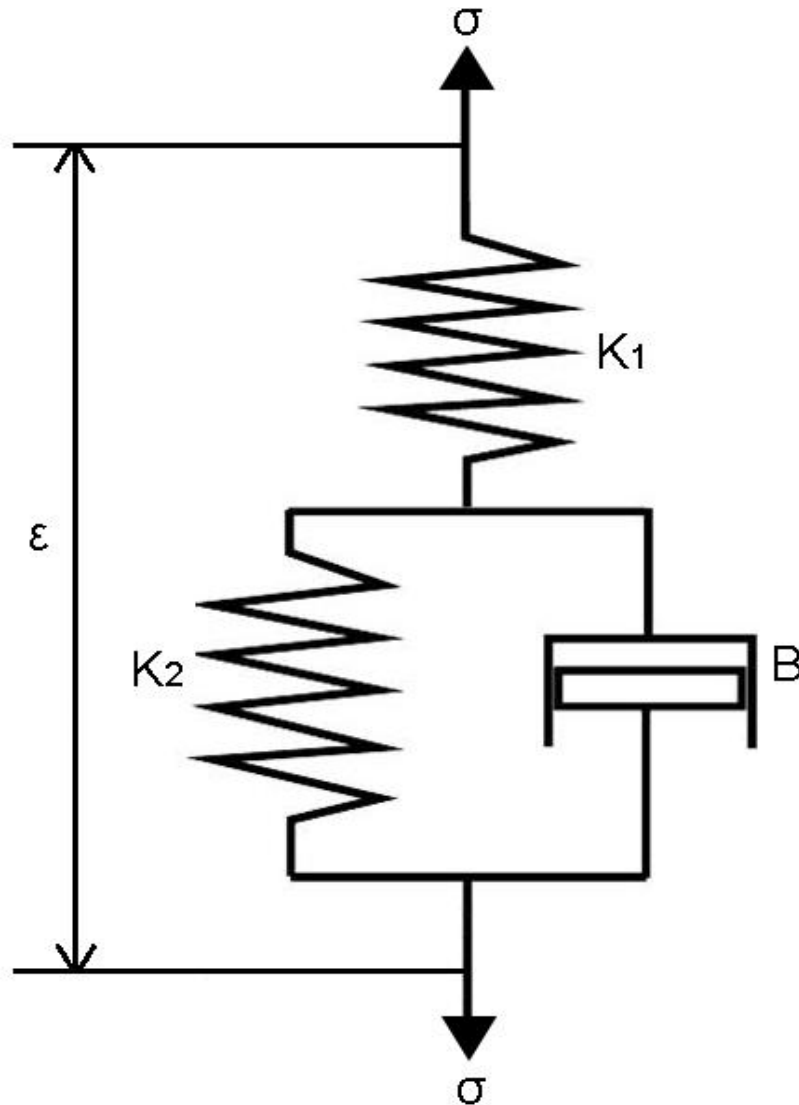


Figure 2-3 : Standard linear solid model

The model is also known as standard linear solid model and it has been used to describe material behaviour in creep experiments, where the material was subjected to constant stress and the deformation was measured, and in stress relaxation

experiments, where the material was subjected to constant deformation and stress relaxation being measured. The model is a combination of spring and dashpot and it is showed in Figure 2-3 and described by:

Equation 2-1 (Fung, 1993)

$$(K_1 + K_2)\sigma + B\dot{\sigma} = K_1K_2\epsilon + K_1B\dot{\epsilon}$$

Where

K_1, K_2 = stiffness coefficients of the springs;

B = damping coefficient of the dashpot;

σ = stress; $\dot{\sigma}$ = first derivate of the stress;

ϵ = strain; $\dot{\epsilon}$ = first derivate of the strain;

Creep recovery experiments have been performed in-vitro in animal (Hult et al., 1995) and in human using human lumbar (Keller et al., 1987, White and Panjabi, 1990, Burns and Kaleps, 1980) and sacral segments (White and Panjabi, 1990).

In the experiment of Keller, Spengler et al. (1987) viscoelastic properties of the T11-L5 spinal segment were evaluated applying a constant load for 30 minutes while compressive load and axial displacement were recorded and used to derive the parameter of the solid model. Different sections of the segment were evaluated and the average error between the experiment date and the data from the model was < 1%.

In-vitro study used creep experiment to evaluate viscoelastic behaviour of the muscle-tendon unit (Taylor et al., 1990) and the data were used to derive stress/strain plots. Kurutz (2006) performed an in-vivo creep recovery experiment in order to evaluate in-vivo viscoelastic parameters of the lumbar spine. Time related elongations of L3-4, L4-5 and L5-S1 lumbar segments were measured using subaquial ultrasound measuring method during 20 min long traction hydrotherapy, where cervical suspension was

applied to participants in a pool. Elongation data were then used to derive viscoelastic coefficients using the three parameters model.

2.6.1.3 Free oscillations

Free oscillations experiments have been used to derive viscoelastic parameters of different human body parts (Ditroilo et al., 2011b, Granata et al., 2004).

This method used the principle that when a system is perturbed and displaced from its equilibrium, the system will start to oscillate to restore the equilibrium. Generally the perturbation may be applied as a displacement or a force, under different conditions, e.g. step input, and the oscillations rapidly decrease according with the viscoelastic properties of the material tested. The pattern of the oscillations can be described by second order under dampening that, in turn, can be represented by a spring-dashpot-mass model. The viscoelastic parameters are derived using natural frequency and damping ratio that are calculated using characteristics of the damped oscillation curve.

The following example, from the work of Walshe, Wilson et al. (1996), shows how musculotendinous viscoelastic parameters for the lower body were obtained through this method. Participants were sitting in a bench generally used for leg press exercise and feet on footplates where a load cell was applied; an external load was applied and the response recorded by the load cell. Oscillatory response showed to be similar to a damped oscillation and the pattern was plotted (Figure 2-4). Damped natural frequency (f) was derived from the graphs as the inverse of the period T between successive peaks and the damping ratio (ζ) plotting the natural log of peak forces against time and obtaining the slope of the line. Damped natural frequency and damping ratio were then used in Equation 2-2 to derive natural frequency (f_n).

Equation 2-2

$$f_n = \left[\frac{f^2}{(1 - s^2)} \right]^{1/2}$$

Where f_n = natural frequency; f = damped natural frequency; s = damping ratio;

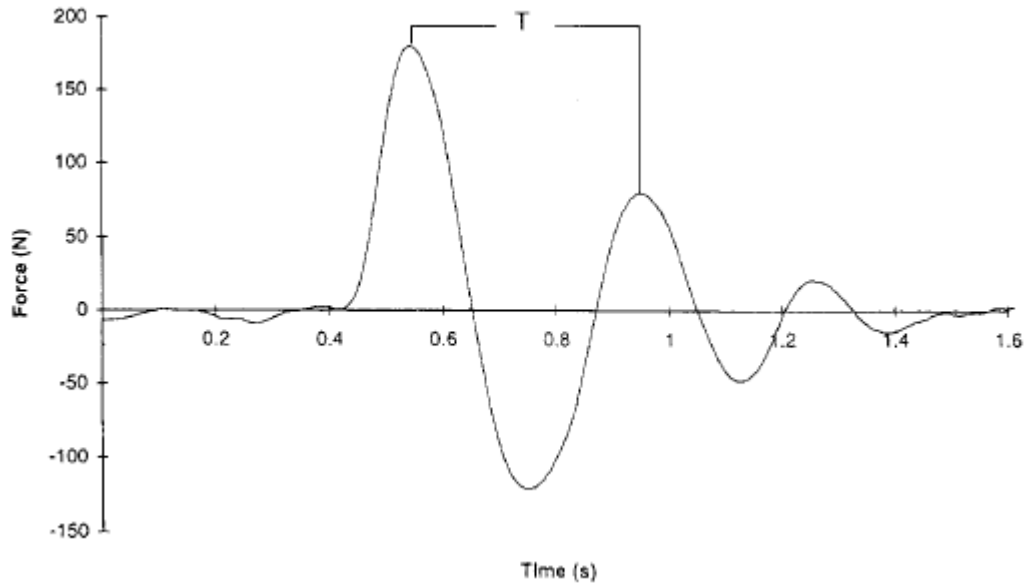


Figure 2-4 : Force pattern of the lower limb from free oscillation experiment (Walshe, Wilson et al. 1996)

The damping (c) coefficient (Ns/m) for a system with mass (m) was calculated with the following equation:

Equation 2-3 (Walshe, Wilson et al. 1996)

$$c = 4\pi m f_n s$$

Stiffness (k) coefficient (N/m) was calculated with the following equation:

Equation 2-4 (Walshe, Wilson et al. 1996)

$$k = 4m f^2 \pi^2 + \frac{c^2}{4m}$$

The test was repeated after 2 days from the first test and reliability was evaluated comparing the stiffness values of the two tests. No significant difference was found between the values from the 2 different days; furthermore values of intraclass correlation coefficient (0.94) and the coefficient of variance (8%) demonstrated the reliability of this test. This method has been used in different studies to evaluate viscoelastic properties of the lower limbs (Ditroilo et al., 2011a, Ditroilo et al., 2011b, Granata et al., 2002, Granata et al., 2004).

2.6.1.4 Mechanical rotational and translational system

A body that is performing a movement can be described using equations that relates characteristics of the body as mass and viscoelastic properties with the motion performed and the force applied to the body. A body rotating can be modelled as a mechanical rotational system using three elements: moment of inertia of the body, a spring and a dashpot. When a torque is applied a body, it starts to rotate and angular displacement, velocity and acceleration can be measured. The body rotating can be represented by a mechanical network (Figure 2-5). Each elements of the network produce a reaction due to the applied torque. The reaction torque produced by the mass of the body equals the moment of inertia of the body times its angular acceleration. The effect of the torque on the spring produces a reaction torque that is proportional to the angular displacement of the body times the stiffness coefficient of the spring. The reaction torque produced by the dashpot is proportional to the angular velocity of the body times the damping coefficient of the dashpot.

Using this mechanical network the following second order linear equation can be derived and it represents the model (D'Azzo and Houpis, 1988):

Equation 2-5

$$T(t) = K\theta(t) + B\dot{\theta}(t) + J\ddot{\theta}(t)$$

Where:

$T(t)$ = torque applied; K = stiffness coefficient;

B = damping coefficient; J = moment of inertia;

θ = angular displacement; $\dot{\theta}$ = angular velocity; $\ddot{\theta}$ = angular acceleration;

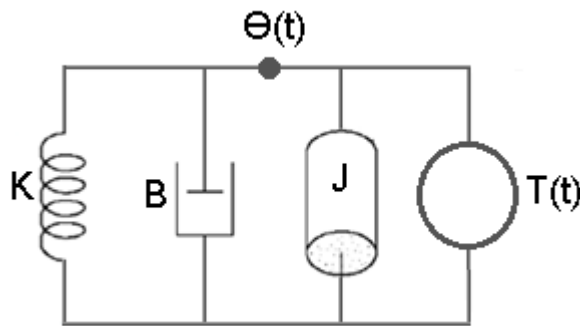


Figure 2-5: mechanical network representing a mechanical rotational system

Similar to this system is the mechanical translational system where the angular parameters (displacement, velocity and acceleration) are substituted by translational parameters, inertia of the body by its mass and torque applied by force (D'Azzo and Houpis, 1988). Both the systems have been used in order to evaluate viscoelastic parameters of human joints. Zhang et al. (2000) delivered a perturbation to the upper limb and the viscoelastic properties of the glenohumeral joint in the abduction axis were derived using data of the applied torque and angular motion of the joint.

Similar experiments were used to evaluate viscoelastic properties of the elbow (Bennett et al., 1992) and ankle joints (Loram et al., 2001).

Trunk viscoelastic properties have been evaluated using the mechanical translational system, applying an external force and evaluating trunk linear displacement in the

sagittal plane in standing (Gardner-Morse and Stokes, 2001, Moorhouse and Granata, 2005) and semi-seated position (Hodges et al., 2009).

In the work of Hodges, van den Hoorn et al. (2009) a participant sat in a semi-seated position with pelvis fixed; an harness, positioned on the shoulder, had weights attached at the front and back via cable as shown in Figure 2-6; two different test were performed, firstly the weight on the front was released without warning and trunk displacement and force vector were recorded. Similar test was performed but the back weight was released and force and displacement recorded.

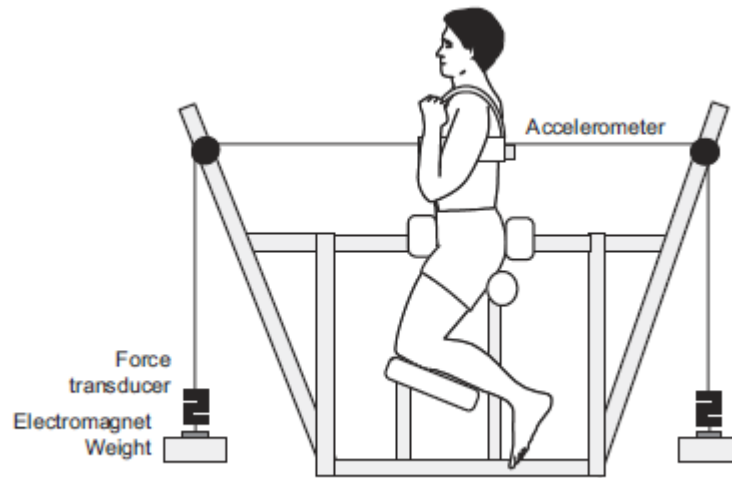


Figure 2-6: Experimental setup for trunk viscoelastic properties evaluation experiment (Hodges, van der Hoorn et al. 2009).

The data acquired were fitted in the second linear order model that described a mechanical translation system:

Equation 2-6

$$F(t) = K\theta(t) + B\dot{\theta}(t) + M\ddot{\theta}(t)$$

Where:

$F(t)$ = force applied; K = stiffness coefficient; B = damping coefficient; M = mass;

θ = angular displacement; $\dot{\theta}$ = angular velocity; $\ddot{\theta}$ = angular acceleration;

The validity of the modelling was assessed by calculating RMS error and coefficient of determination (R^2) between measured and modelled displacement. The RMS error was < 1 mm and the R^2 was >0.98 for both the tests, indicating that the model accurately describe the test. An example of the modelled and experimental data is shown in Figure 2-7, where the displacement measured during the experiment is plotted with the displacement calculated through the second order linear model.

Effective trunk mass was obtained and the results were used to validate the model: the effective mass trunk showed correlation with total body mass and an addition of mass was estimated accurately by the model (participant with and without an extra weight were performing the same protocol and the error in mass estimation was $> 8\%$). 5 subjects were tested again after 2 days for reliability and the intraclass correlation coefficient for damping, stiffness and mass in forward perturbation was found >0.65 .

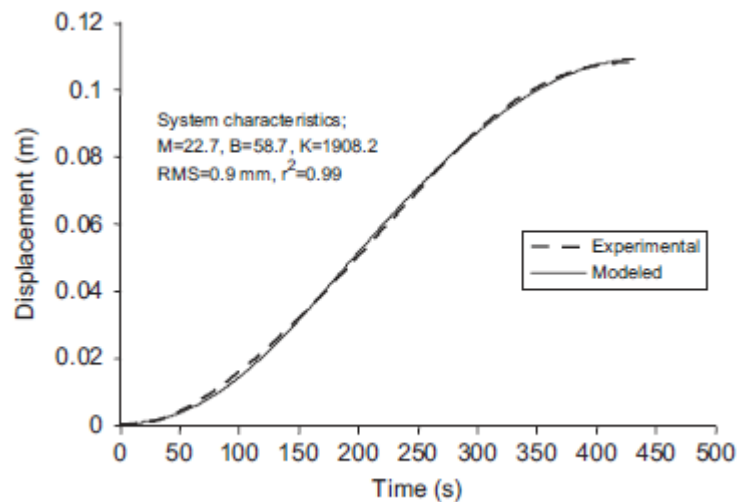


Figure 2-7: Experimental and modelled displacement data for one participant in the trunk viscoelastic experiment; trunk effective mass (M), damping (B) and stiffness (K) are showed as well as RMS of the error and coefficient of determination (R^2) (Hodges, van der Hoorn et al. 2009).

Viscoelastic properties of the trunk was evaluated in similar studies using mechanical rotational system, collecting trunk angular displacement and torque instead of linear

displacement and force applied with participants in lying (Brown and McGill, 2009) and semi-seated (Cholewicki et al., 2000b) position.

2.6.1.5 Comparison between viscoelastic models

In this section different models have been developed to evaluate viscoelastic properties of the living tissue. The aim of this research is to assess the viscoelastic properties of the trunk while the participant is performing a balancing task in sitting posture.

The free oscillations work when the oscillating part being measured behaves as an under-damped oscillator. The three parameters solid model demonstrates the response of the material to prolonged stress or strain in creep recovery and stress relaxation experiments, especially in in-vitro experiments.

The mechanical rotational system showed to be able to model different joints in different configurations and with the participant performing different tasks.

In this research participants will try to balance on a custom made swinging chair. Participants will rotate their trunk to find the balanced and the angular displacement, along with the trunk moment will be measured. The mechanical rotational system model will be used to evaluate the viscoelastic parameters of the trunk for the participants performing the balancing task.

2.6.2 Viscoelastic properties and LBP

LBP has been related with spinal instability and mechanical alteration of the lumbar spine as been found to be involved in this mechanism (Panjabi, 2003). In-vitro studies showed intrinsic alteration of the viscoelastic properties of the intervertebral discs in degenerated discs; in particular stiffness and damping coefficients, derived in a creep experiment of lumbar spine segment specimens, were found decreased for

degenerated discs (Keller et al., 1987) and ,as consequences, these segments were found less stable, with an higher creep rate than less degenerated segments.

Alterations in hip joint and trunk viscoelastic properties have been evaluated in LBP subjects in various in-vivo studies. Regarding the hip joint, Tafazzoli and Lamontagne (1996) used a isokinetic device that raised the leg at slow and constant velocity to calculate passive elastic moment and the stiffness coefficient of the hip joint in healthy and LBP subjects. Damping coefficient was calculated using the suspension method, where the lower limb was hanging in another device provided with a spring support system and the springs were stretched in order to produce an oscillation. The results showed that stiffness and passive moment were increased in LBP participants while no differences were found in the damping coefficient. Increased stiffness of the hip joint and decreased trunk flexibility (evaluated with a flexometer in a bending forward test) were successfully used to distinguish symptomatic from healthy subjects, implying that decreased flexibility and tightness of the hamstring muscles may be associated with LBP (Tafazzoli and Lamontagne, 1996).

Postero-anterior lumbar spine stiffness has been investigated and compared in LBP and healthy participants by Latimer, Lee et al. (1996b) that used a device that delivered a force to the subject at L3 level while subject is lying on a table in prone position, and stiffness was calculated as function of force and displacement as described earlier in this section. Results showed that LBP was related with the increasing of stiffness in the trunk. Moreover LBP subjects were tested while they were experiencing pain (test 1) and after that the pain was resolved by more than 80% (test2) and results showed a significant decrease in the stiffness in the second test (Latimer et al., 1996b), highlighting the association between increased stiffness and LBP.

Colloca and Keller (2001) used an improved version of the device used by Latimer, Lee et al. (1996b), in order to evaluate the posteroanterior stiffness at different levels of the

trunk (left and right PSIS, sacral base left and right, S1, L5, L4, L2, T12 and T8 for the spinous and transverse processes) with the participants in prone position, and their results were in accordance with previous study, finding that LBP subjects with high recurrence of pain had increased stiffness in the spinous processes (stiffness was measured for each vertebra and mean of all vertebrae calculated) in comparison to healthy subjects or subjects with less frequent LBP.

Hodges, van den Hoorn et al. (2009) investigated trunk viscoelastic properties modelling force and displacement parameters from a trunk perturbation experiment with a second linear order system as described earlier in this chapter. Trunk stiffness coefficient was found significantly increased (mean of the stiffness 1641 N/mm for healthy and 1997 N/mm for LBP participants with p-value <0.05) and damping decreased (mean of the damping 55 Ns/mm for healthy and 17 Ns/mm for LBP participants with p-value <0.05) for LBP participants after they experienced a forward perturbation. Similar results were found for backward perturbation, but the increasing in stiffness mean for LBP patients was not significant. The researchers suggested that the increased stiffness may be due to increased trunk muscle activity to protect spinal structures and a mechanism to compensate the reduced damping coefficient that may be due to physiological changes in passive structures (Hodges et al., 2009).

2.7 Trunk muscular activity

2.7.1 EMG analysis of the trunk muscles

The activity of different muscles of the human body has been monitored by electromyography (EMG).

Different studies evaluated the muscular activity of the muscles involved in the trunk stabilisation and motion. In previous studies muscles have been monitored using superficial electrodes; these electrodes were placed on the left and right side of the body on the erector spinae, rectus abdominis, internal and external obliques (Krajcarski et al., 1999, Preuss et al., 2005, Reeves et al., 2006, Thomas and Lee, 2000). Other muscles such as the transversus abdominis or quadratus lumborum are impossible to be monitored with surface EMG because they are located inside the abdomen below other muscles or organs they can be monitored with intramuscular needles, but they are invasive and they need to be placed by a trained specialist.

Amplitude of EMG signals from different subjects has been compared using maximal voluntary contraction (MVC) as normalisation technique. In this method, the MVC for each muscle is measured using a specific task that allows the largest amplitude of the myoelectric activity for that particular muscle to be produced. The EMG activity of each muscle during the experimental task conducted is then reported as percentage of the MVC. This allows comparison between subjects and trials (McGill, 1991).

This method has been used by several studies that evaluated the muscular activities of the trunk muscles during assessment of spinal stability in semi-seated position (Cholewicki et al., 2000b, Brown et al., 2006), in standing position (Krajcarski et al., 1999) and during unstable sitting (Preuss et al., 2005, Reeves et al., 2006).

Other methods used for data normalisation are the mean dynamic method and the peak dynamic method that use data from the same task. These methods are effective in

improving group homogeneity but they do not allow amplitude comparison between different muscles and subjects (Burden et al., 2003, Yang and Winter, 1984).

Other studies have investigated the muscle on-set and off-set time which showed when a certain muscle was considered active during certain tasks (Forssberg and Hirschfeld, 1994, Gilleard et al., 1998, Stokes et al., 2000). A threshold method is used to establish the on-set/off-set times. The threshold was generally taken from resting data, calculating the average of the resting signal and adding to it three times its standard deviation. The EMG signal of the experiment was then analysed and when the signal exceeded this threshold for a certain time (generally 50 ms) an on-set time was recorded. The off-set was detected using the same threshold but analysing the EMG signal reversing the time i.e. from the end of the experiment till the starting point. In this manner muscle latency was calculated as the time difference between starting of the task and first on-set of the muscle (Leinonen et al., 2001). Moreover activity and co-contraction of the muscles was quantified as percentage of the total length of the experiment (Radebold et al., 2000).

The frequency content of the EMG signal has been used to evaluate muscular fatigue in the back muscles (Dolan et al., 1995).

In order to detect side asymmetry between muscles from the same group (e.g. left and right external obliques), correlation coefficient is calculated using the signal acquired from the left and the right muscles of a certain group (Nelson-Wong et al., 2009).

2.7.2 Muscle spasm

There is not a clear definition for muscle spasm; Mense, Simons et al. (2001) suggested that muscle spasms may be defined as *“the electromyographic (EMG) activity that is not under voluntary control and is not dependent upon posture. It may or may not be painful”* where painful muscle spasms were often reported as muscle cramps. Other

authors considered any abnormal increasing in muscle tension as muscle spasms, but in this case the definition could overlap the definition of muscle tone. In some cases also painful EMG-free tension of the muscle was defined as muscle spasm or cramp, but the medical definition of this is muscular contracture. According with the definition, muscle spasms can be monitored and quantified using surface EMG in subjects during resting (Mense et al., 2001).

Two models have been suggested to explain the effect of pain on muscular activity. The pain-spasms-pain model, based on that theory, was built in order to explain the relation between changes in muscular activity and pain. In this model, pain causes a muscular spasm (increased muscular activity), that in turn leads to more pain (Travell et al., 1942). The pain adaptation model, suggests that the pain stimulus is involved in decreasing activation of the agonistic muscles and in increasing the activation of antagonistic activity (Lund et al., 1991).

The review conducted by Van Dieën et al. (2003) indicated that neither model was well supported by the literature, with some studies revealing a decrease in lumbar erector spinae activity, in accordance with the pain adaption model; while others found increased activity, in agreement with the pain-spasms-pain model. Moreover side asymmetry and irregular patterns were found in the lumbar muscles showing effects of the pain on the motor control. The alteration of the muscular activity could be an adaptation strategy to pain but the proposed pain adaptation model also had limitations (van Dieën et al., 2003). Length of symptoms, different origin of the pain (Arena et al., 1989) and task involved in the tests (Sherman, 1985) influenced the results of the studies. It was proposed in the review that the alterations in the muscular activity may be influenced by different factors and may be also a strategy to avoid risks of further injuries in the lumbar spine structures. As described by Panjabi (2003), increased muscular activity may be a compensatory response to decreased stability: stiffness of

the spine could be enhanced contracting the muscles; furthermore the stiffening of the trunk achieved with the co-contraction of the muscle may compensate for the decreased ability of reacting to perturbation, reducing the trunk range of motion (ROM). Furthermore alterations in the muscular activity can decrease risks of unwanted motion and further damage but they are also related to negative consequences such as increased spinal load, fatigue and possible injuries in the spinal muscles (Granata and Marras, 2000, van Dieen et al., 2003).

Williams et al. (2010) reviewed studies with experimentally induced pain and they found kinematic alteration in LBP patients due to changes in muscular activity. Central nervous system (CNS) showed increased time in motion control and modifying of muscular activity in order to reduce the risk of pain provocation, but it could be also possible that pain could induce muscular activity and kinematic alteration that can lead to the decreasing in spine health. Further studies are needed to understand if and how pain drive this kinematic and muscular changes (Williams et al., 2010).

There is not clear evidence of the role and the presence of muscle spasms in people with musculoskeletal pain in general (Johnson, 1989). As described previously, the increased activity of the muscle in LBP patients can be related to altered trunk kinematics due to pain rather than the pain itself (Williams et al., 2010).

As a result, other biomechanical variables need to be taken in account in order to fully understand alterations in biomechanical behaviour of the trunk due to LBP and to clarify the mechanism underlying muscular activity changes in LBP patients. This will help to address and rationalise the work of the clinicians to restore the altered biomechanical behaviour of the trunk in LBP patients.

2.7.3 Muscular activity and LBP

Altered activity of the trunk muscles has been found in people who experience LBP but, as mentioned above, the pain-spasm-pain and pain-adaptation model have not been fully supported by literature.

Another possible explanation was presented by Panjabi (1992), where an alteration of the intrinsic stiffness of the passive structures due to injury can lead to increased muscular activity as a compensatory response in order to sustain trunk stability.

Several EMG studies investigated the mechanism behind the EMG activity alterations due to LBP evaluating different aspects of the activity of the muscles involved in the trunk posture and motion. Regarding signal amplitude, average and root mean square (RMS) of the normalised signal from different trunk muscles were used to quantify and compare muscular activity between symptomatic and asymptomatic people performing different tasks.

Ambroz, Scott et al. (2000) measured the EMG activity of the trunk placing four electrodes symmetrically in the right and left lumbar paraspinal region (from L1 to L5) and measuring activity while a participant was in standing position. Results showed a significant increase of the muscular activity, with the muscular activity average of the LBP participants group threefold higher than in healthy subjects. Similar results were found by Sihvonen, Partanen et al. (1991) who found increased RMS of the paraspinal muscle activity calculated using intramuscular EMG in LBP patients. Silfies et al. (2005) studied the behaviour of the trunk muscle in a dynamic test where participants started in a standing position with the trunk extended to 15° and then performing a trunk flexion to reach 15° of forward bending; the experiment was repeated with the participants holding a load of five pounds. Results from different muscles showed that for two different experimental setups (with and without holding the load) there was a significant increase in RMS of the EMG signal for external oblique and rectus abdominis muscles

in subjects with chronic LBP while no differences were found in the internal obliques, erector spinae and lumbar multifidus.

Research involving low-level isometric rotation in standing position found higher normalised RMS of the EMG signal for the gluteus maximus, hamstring and erector spinae muscles in chronic LBP subjects that was explained as a strategy to increase spine stability and decrease the risks of further damage (Pirouzi et al., 2006).

During comfortable walking the normalised average of the EMG signal from lumbar erector spinae activity was found higher in chronic LBP, moreover in the same study during different gait speeds the erector spinae also showed a decreased coordination of its activity with kinematics data when compared with healthy subjects (Lamoth et al., 2006). These studies showed that LBP exhibited alterations in the amplitude of the signal, in particular higher activity in terms of average and RMS of the EMG signals were found in people who experienced back pain.

Signal amplitude has been also analysed using principal component analysis, using healthy subjects as reference model, to evaluate possible difference in the signal pattern due to pain. EMG signal recorded from healthy subjects were used to build principal component (PC) models, from which distance measures were calculated and compared to the model generated from LBP data. Data showed that the model was sensitive to the task (e.g. loaded or not loaded lifting) but poorly sensitive to the participant condition (Larivière et al., 2000b), demonstrating poor suitability for evaluating LBP effects.

Another important factor that has been used to investigate muscle behaviour is time: duration of contraction and time delay between perturbation and muscle response gave important information about alterations in activity and muscle recruitment.

Leinonen et al. (2001) evaluated the muscle response for the multifidus and erector spinae muscles in healthy and chronic LBP participants while they were standing with a

box on their hand and a load was dropped into it; the experiment was replicated with participants having the eyes closed. Duration of the activity was calculated as time differences between onset and offset of the muscles and latency time was calculated as the time difference between the instant when the load hit the box and the onset of the muscles. No differences were found in the duration of the activity between groups. Healthy participants exhibited a faster reaction time when the eyes were opened, while for LBP no differences were found between eyes closed and opened and researchers concluded that there was evidence of impaired feed-forward control of the lumbar muscles due to pain.

Reaction time was also evaluated in other studies involving sudden trunk loading: Radebold, Cholewicki et al. (2001) measured the response for trunk muscles (rectus abdominis, latissimus dorsi, erector spinae, external and internal oblique) to sudden perturbation. In the study a participant sat in a semi-seated position performing an isometric trunk exertions and the resistance load was suddenly released. Results showed that response time was on average 15 ms faster in healthy subjects implying an effect of LBP on the motor control mechanism.

As mentioned above LBP has been related to stiffening of the trunk in order to compensate with decreased intrinsic spine stiffness and to avoid dangerous movements (Panjabi, 1992) and the contracting simultaneously trunk agonist and antagonist muscles has been proposed as a method to achieve this. For this reason researchers tried also to quantify the co-contraction in the trunk muscle using agonist/antagonist activity ratio or time length of the co-contraction.

Van Dieen et al. (2003) found that the ratio of the EMG amplitudes in the antagonist over agonist muscles were increased in LBP patients in a test where participants performed slow trunk motion in a semi-seated position (e.g. sagittal, frontal and

transversal plane movements). The increased activity was concluded to be a compensatory strategy in response to decreased spine stability due to pain.

In the study of Radelbold, Cholewicki et al. (2000) participants experienced a sudden perturbation in a semi-seated position and co-contraction was evaluated as the temporal duration of the simultaneous activity of the agonist and antagonist muscles. Results showed that LBP subjects exhibited contemporaneous activity while in healthy subjects muscles were switching on and off alternatively; these results were in agreement with the findings of the studies mentioned previously.

Also in the study from Silfies et al. (2005) mentioned above, co-contraction between flexors and extensors (the ratio between the sum of the normalised flexors activation and the sum of the extensor normalised activation), was evaluated. In contrast with previous studies, no differences were found due to LBP, but while for the other studies the data refer to all the length of the trial in this research the data refers to a specific point of time i.e. when the participants, during flexion, reached the 0 degrees position.

2.8 Trunk kinetics

2.8.1 Kinetics evaluation for trunk motion

Kinematic gives useful information about the range of motion and the mobility of different joints such as hip and lumbar spine during a certain tasks, but it is also important to evaluate kinetic variables such as joint moment, power, and loading characteristics in order to evaluate how forces are acting in the spine and if a certain condition could alter these variables. Furthermore kinetic studies can also evaluate spinal loading during different tasks, suggesting strategies to reduce dangerous stress that a certain task may create on the spine. In-vitro studies, using creep tests, showed stress concentration in the intervertebral discs when constant load is applied, that may lead to structural degeneration and disruption of the spinal vertebrae (Adams et al., 1996). Furthermore Gallagher, Marras et al. (2005) in their in-vitro study investigated how the spinal loads changed with lifting a 9 kg weight with the torso bended at different angles; increasing in torso angle was associated with increasing in spinal loading and in decreasing in time for fatigue failure of the lumbosacral structures, suggesting that common activity as weight lifting may be potentially dangerous and recommendation should be given to decrease risk of injuries.

Spinal loading during different activities has been evaluated in several in-vivo studies while participants were performing different tasks in order to quantify the mechanical stress undergone by the spinal structures. In these studies trunk kinetic and loading has been calculated merging trunk kinematic and forces and moment data using inverse dynamic equation during different tasks. Trunk loading characteristics were evaluated by Marras et al. (2001) while participants were performing controlled and free dynamic exertions. During the tasks participants were in standing position on a force platform with pelvis fixed; participants were initially asked to reach a predetermine force exertion

level and then to control sagittal extension; thereafter they were asked to perform a free dynamic lifts using different weights. The loading at level of L5/S1 joint centre were evaluated for the three planes of motion in order to have medio-lateral share, antero-posterior shear and compression loading using a EMG assisted model (Marras and Granata, 1997a). Furthermore sagittal trunk moment was derived using motion and force platform data. During both the tasks compressive loading was higher than lateral shear and anterior shear loadings, and lateral shear loading was higher than anterior shear loading, describing how the stress was divided in the three planes of motion.

In similar studies loading of the spine during lateral bending motion (Marras and Granata, 1997b) and axial twisting of the spine (Marras and Granata, 1995) was investigated. In the lateral bending experiment the participant was standing over a force platform with the pelvis fixed and asked to bend to left and right side with a weight on his hand. The experiment was performed at different trunk speeds. The results showed that the compressive and the lateral shares loading increased together with the increasing of trunk velocity; moreover co-contraction of trunk muscles increased with trunk speed, that is in agreement with increased spine loading (Marras and Granata, 1997b).

Results about lateral bending were in agreement with results from the axial twisting of the spine, where the participant was performing axial twisting exertion at different speeds and load exertions (Marras and Granata, 1995). Spinal loading was found to increase with velocity, exertion load and twist angle that was also in agreement with studies that related risks of back injuries with these variables (Bigos et al., 1986, Schaffer and Statistics, 1982).

Lifting is a common task that may involve risks for the health of the human spine (Edlich et al., 2005). Loading and kinetics data of the spine during this kind of task are important to understand the risks involved and to reduce them.

Schipplein, Trafimow et al. (1990) evaluated moments at L5/S1, hip and knee joint levels while participants were lifting weights with self selected technique and speed. Results showed that with increasing weight, the loading sharing between different joints changed, in particular L5/S1 moment increased and knee moment decreased; furthermore knee joint angular velocity was increased with the weight. Lifting tasks were also found to be dependent on velocity: back and arm lifting strengths were found different at different stage of the lift and inversely related with the speed of the task (Kumar et al., 1988); furthermore compressive and shear loading at L5/S1 were increased with increasing lifting speed and weight (Granata and Marras, 1995a). Kinetic and loading of the spine during lifting were found to be related with factors such as working place, techniques and shape of the lifted objects. Granata, Marras et al. (1999) found high variability for kinetics, kinematics and loading of the spine variables evaluating people performing the same tasks but in different environment condition (e.g. lifting with asymmetry in the leg position) or with different experience in the job. Differences were also found in a study that was evaluating lowering and lifting tasks: higher compressive loading while lesser anterior-posterior shear loading was found in lowering than in lifting (Davis et al., 1998). These differences were explained by the researchers as a result of different techniques used to hold the weight in lifting and lowering: trunk moment in lowering was found increased than in lifting, implying that the weight was held further away from the body than in lifting (Davis et al., 1998). Shape of the object to be lifted has an effect on the spinal loading: participants lifting boxes with same weight but different shape were found to have difference in compressive loading in the L5/S1 joint, especially larger boxes showed to increase the compressive loading in the participant's spine (Freivalds et al., 1984). Kinetics variables were also used to evaluate fatigue during repetitive lifting tasks by Sparto, Parnianpour et al. (1997). Participants were standing on a force platform and

asked to bend forward and to lift a weight from the floor while motion was recorded; knee, hip and lumbosacral joint average moments were derived and compared. In the beginning, when the participants were not fatigued, hip moment was higher (average: 210 ± 39 Nm) than lumbosacral moment (average: 188 ± 39 Nm) and knee moment (average: 51 ± 54 Nm). The moments on the hip (average: 179 ± 26 Nm) and lumbosacral joints (average: 159 ± 24 Nm) were found to decrease significantly after the participants were fatigued, as well as lifting force. However joint relative work did not change, showing that the relative contribution to the tasks of each joint was not influenced by fatigue.

Spinal loading during walking was evaluated by Callaghan, Patla et al. (1999) and results demonstrated that compressive, anterior-posterior shear and lateral shear loadings at L4/L5 joint level were below values of spinal loadings caused by other activities used in trunk rehabilitation (e.g. back extensor exercise) (McGill, 1998), suggesting that walking may be used as rehabilitation exercise. The spinal loading was also found to be related with walking speed and arm swing: increasing spinal forces was found when walking was reduced and the arm motion restricted, showing that a more rigid trunk during walking may be related with increasing stress in the spine.

Moment, velocity and power at hip and lumbospinal joints have been analysed while performing sit-to-stand and stand-to-sit in symptomatic and asymptomatic subjects (Shum et al., 2007b). Trunk motion was monitored using electromagnetic tracking device and force platform measured ground reaction force. Inverse dynamic equations were used to derive moments at hip and lumbar spine level. Hip and lumbar spine joint powers were derived through joint moments and angular velocities. Sit-to-stand task was analysed and results showed that the task was formed by 2 phases, first and second phase, divided by the transition phase. The first phase started when the thighs were off the chair; in this phase lumbar spine and hip flexed, reaching quickly their peak

and then extended. Extension muscle moments were developed throughout the task. The power for both joints in this phase was negative, implying that hip and spine extensors muscle were working eccentrically. Moment, velocity and motion for both joints were minimal in the other planes of motion. Transition phase started when power was zero and the muscles were switching from eccentric to concentric action. The second phase started when the power was positive and the muscles were working concentrically to extend the trunk to the final standing position. In stand-to-sit action powers and moments exhibited similar pattern to that of sit-to-stand. Extension moments were throughout the task and muscles worked eccentrically (during trunk and spine flexion) and then eccentrically (when the participant was reaching the sitting posture). Moment, power and motion were small for both joints in frontal and transverse plane of motion.

2.8.2 Kinetics and LBP

Cumulative and peak spinal loading, increasing in trunk velocity and increasing in external forces (e.g. weight of lifted objects) have been found to increase risk of develop LBP in working environment (Kumar, 1990, Norman et al., 1998).

Alterations in kinetic variables were found in LBP patients while performing different tasks. Trunk loading characterises were evaluated by Marras, Davis et al. (2001) while participants were performing controlled and free dynamic exertions. In the experiment motion, moment, spinal loading and EMG of trunk muscles were measured and compared between healthy and LBP participants. Spine compression and lateral shear were found to increase in LBP subjects. In free dynamic range task LBP participants exhibited increased spinal loading and decreased motion and velocity of the trunk, but the loading differences disappeared when they were normalized for unit of moment exposure, implying that kinematic trunk alterations were a compensatory strategy in

order to reduce effective spinal loading. In both conditions electrical activity of the trunk muscle and body mass were found significantly higher in symptomatic subjects, suggesting that they played a primary role in increasing absolute spinal load.

Lift origin locations showed to influence the spinal loading, in particular the difference in the spinal loading between LBP and healthy subjects was found to be higher for waist and shoulder lift origin when compared with knee lift origin. These findings may help clinician to develop guideline for decrease potential risk for LBP in working environment (Marras et al., 2004).

LBP alterations in kinetics variables were evaluated by Shum, Crosbie et al. (2007b) in the study about sit-to-stand and stand-to-sit tasks described earlier in this section. Data between 20 healthy, 20 LBP and 20 LBP with SLR alteration participants were evaluated and compared. In both the tasks analysis showed decreased moment of the hip and lumbospinal joints in the sagittal plane, which was compensated with increased moments in the transverse and frontal planes of motion in LBP participants. Hip and spine joints angular velocities were also decreased in symptomatic group. Due to differences in joint moments and velocities around the hip and spine, power curve pattern for both joints were altered in LBP participants, in particular the peak power generation and absorption were decreased in symptomatic subjects. Differences were believed to be a strategy in order to decrease pain and the risk of damage in the spine tissue.

2.9 Thoracolumbar curvature

2.9.1 Measurement methods

Different techniques have been used to assess the static thoracolumbar curvature. Lateral radiographic images have been widely used to measure kyphosis and lordosis angle (Jackson and McManus, 1994, Tsuji et al., 2001) evaluating images of the trunk and calculating posture angle through visual or digital analysis; the method showed to be valid and reliable, but it carries disadvantages as radiation risks and high cost for the apparatus.

In order to reduce risks connected with radiation, photographic method was used to evaluate spinal curvature, attaching markers on the skin overlying relevant spinal processes (e.g. L1; S1; T1) and deriving angles using tangents at these points. The method showed reliability but the data showed high variability when different operators evaluated the same participant through visual analysis.

Flexi curve and khyphometers are easily operated, inexpensive method to evaluate static parameters from the lumbar spine (Lundon et al., 1998) and they do not carry risks for participants.

Spinal mouse is a device that consists of two rolled wheels and an accelerator; while the wheels are passed on the skin overlying the spinal column, the accelerometer evaluates spinal inclination on the sagittal plane. This method has shown to be a reliable and easily operated but only sagittal plane curvature can be obtained, and there is not much information on how the angles were derived (Mannion et al., 2004).

Singh et al. (2010) used electromagnetic motion system to evaluate the curvature, a wooden probe attached to an electromagnetic sensor was used firstly to digitalise bony landmarks as spinal processes and PSISs. Digitalised points were used to measure kyphosis and lordosis angles, while a three dimensional draw of the thoracolumbar

curvature were obtained passing the wooden probe along the skin underlying the spinal column. The method was found to be highly reliable (0.9 intra-class correlation coefficient for sagittal plane). Data acquired using this method allowed further analysis e.g. frontal plane data analysis for scoliosis evaluation.

In summary flexi curve, khyphometers and electromagnetic motion system showed to be the best option to evaluate thoracolumbar curvature and investigate kyphosis and lordosis angles. In particular electromagnetic motion system showed to be reliable and to give more information about spinal static posture than the other methods. Photographic method are cheap and reliable, but the analysis carries error and it is time consuming. Although radiographic methods showed to be valid and reliable, the use is not recommended for static trunk posture evaluation because of the high risks and the low cost-effectiveness that have been demonstrated.

2.9.2 Relationship between posture and LBP

Alterations in the thoracolumbar curvature have been linked with different spinal disorders and their assessments have been used for clinical decisions (Singh et al., 2010). LBP patients demonstrated to have altered sitting and standing postures when compared with healthy subjects. Radiographic studies showed decreased lumbar lordosis (defined as the angle between L1 and S1) in people affected by LBP (Jackson and McManus, 1994) and also in the elderly (mean age 67.8 ± 5.8 years) population with LBP symptoms (Tsuji et al., 2001) when they were tested in normal upright standing position. Moreover the decreased lumbar lordosis was also significantly correlated with the visual analogue scale (VAS) of pain. No differences were found in thoracic kyphosis between healthy and LBP subjects (Jackson and McManus, 1994) in upright standing position.

Studies found relationship between working place and occurrence of LBP, especially in jobs where people had to sit for long period of time (Lis et al., 2007, Andersson, 1999). Differences were found between LBP and healthy people in the sitting posture while performing different tasks. When participants were asked to sit in their usual manner, LBP subjects exhibited a greater posterior pelvic tilt when compared with healthy subjects by O'Sullivan, Mitchell et al. (2006); furthermore increasing in posterior pelvic tilt was also correlated with decreased back muscle endurance, that in turn, was correlated with increased sitting period and decreased physical activity. Researchers showed that LBP people tend to sit for longer time and to reduce physical activity and this may explain the decreasing in muscular endurance. Moreover LBP subjects sitting posture (with greater pelvic tilt) showed to be more passive, with less use of the trunk muscles, but with an increasing in spinal loading that could lead to worsening of symptoms.

Studies evaluating different types of employments showed that the incidence of the LBP was higher for jobs where the person had to remain seated, as was found by Bovenzi and Zadini (1992) that compared public bus drivers with maintenance workers and they found a significant increasing in the LBP incidence for the drivers. Further studies also showed that incidence of LBP was higher in tractor drivers than in office workers that were able to adjust their sitting posture (Bovenzi and Betta, 1994).

Sitting posture can also be involved in the aggravation of the symptoms because a wrong posture may lead to increase of spinal loading that can cause spinal damage. It has been questioned if postural correction in LBP patients may be beneficial for their conditions; Scannell and McGill (2003) in their study evaluated people with normal, hypolordotic and hyperlordotic lumbar posture in different tasks (standing, sitting and walking). Firstly they calculated a neutral zone (range of lumbar lordosis angle where passive strain of tissue was minimised) and they evaluated if the lordosis angle was in

this range during the activities and then they applied to the participants an exercise programme in order to change subjects postures and to have lumbar lordosis in the neutral zone. Results showed that hyperlordotic participants had more tissue strain in standing while hypolordotic in sitting and normal subjects were always in the neutral zone. The exercise programme showed to effectively decrease the tissue strain because all participants were performing their activity with the lordosis angle in the neutral zone. Researchers pointed out that in sitting hypolordosis had greater tissue strain than hyperlordotic, which implies that postural correction and exercise programme need to be rationalised for different conditions (Scannell and McGill, 2003).

2.10 Clinical management of LBP

2.10.1 General interventions

Different interventions are available for LBP management that are related with the nature of the pain and with the strength of the symptoms; treatments focus on reducing pain and improve function and can be divided as (Airaksinen et al., 2006, Savigny et al., 2009):

- Passive physical treatments: e.g. laser therapy, traction therapy and therapeutic ultrasound therapy;
- Manual therapy: e.g. spinal manipulation, massage;
- Exercise therapy;
- Cognitive behavioural therapy and self management;
- Multidisciplinary interventions: they involve different aspects as medical, physical, vocational and behavioural components and they are developed with the coordination of different professionals (physicians, physiotherapists, psychologists);
- Pharmacological treatments: e.g. antidepressants, muscle relaxants, painkillers;
- Invasive procedures: e.g. injections and nerve blocks, percutaneous electrical nerve stimulation, acupuncture, spinal surgery.

2.10.2 Exercises for LBP management

Abenhaim et al. (2000) defined exercise therapy as *“a series of specific movements with the aim of training or developing the body by a routine practice or as physical training to promote good physical health”*. It is widely used and recommended to improve and manage the condition of LBP patients (Savigny et al., 2009).

There are different types of exercise programmes that are used for LBP treatment, management and rehabilitation; the main types of exercise programmes are as follow:

- Mobilisation exercises: the main aim of these exercises is to restore a normal mobility of the joints (Hertling and Kessler, 2006);
- Stabilization exercise: these exercises are specifically designed to target muscles that are involved in spinal stabilization as multifidus and transversus abdominis that showed localised segmental dysfunction (Hides et al., 2001a). Spinal stabilization exercises aim to retrain patients to control and adapt joint range of motion according with activity and position (Hertling and Kessler, 2006). Sling exercises are particular exercises used for retraining muscles as multifidus and transversus abdominis; in this type of exercising the sling gives an unstable base support and muscles need to increase their activity in order to maintain balance (Unsgaard-Tøndel et al., 2010);
- Strengthening exercises: LBP patients demonstrated decreased strength in the trunk muscle (Reid et al., 1991, Shirado et al., 1992); these exercises aim to enhance strength of the trunk muscles through specific intensive dynamic exercise in order to recover, prevent and reduce disability caused by LBP (Handa et al., 2000, Durstine and Medicine, 2009);
- Postural correction exercises: some LBP patients showed extreme lumbar posture as hypolordosis and hyperlordosis, which may be clinically related to increasing in lumbar passive tissue strain. These exercises aim to correct the lumbar lordosis to a neutral posture that has shown to effectively decrease strain in the trunk in everyday activity as standing and sitting (Scannell and McGill, 2003);
- Aerobics exercises: these exercises were found helpful for lumbar rehabilitation and maintenance after recovery from LBP (Mannion et al., 2001, McGill, 1998,

Sculco et al., 2001). Moreover aerobic exercises help improving the psychological condition of the patient, reducing depression and increasing pain tolerance (Nutter, 1988). During aerobic exercises as walking or cycling, loading on the spine is minimised but the trunk musculature is activated showing that these activities can train the lumbar spine without excessive stress on the tissues (Mannion et al., 2001, McGill, 1998, Sculco et al., 2001);

- Williams flexion exercises: In his studies Williams suggested that during standing posture human beings deform the vertebral column, moving the body weight more to the posterior aspect of the vertebrae. This implied that standing lumbar lordotic posture in every human being gives excessive strain to lumbar tissue and that is the cause of LBP. The aim of his exercise protocol is to decrease the lumbar lordosis at minimum to reduce loading on the posterior aspect of the vertebrae (Williams, 1974, Ponte et al., 1984);
- McKenzie therapy: This is based on the McKenzie theory that suggests that spinal pain can be attributed to mechanical deformation of the nucleus pulposus of the discs after alteration of their position due to lifestyle that generally tend to decrease lumbar postural extension. The extension biased exercising programme is the most common programme developed to restore or maintain lumbar lordosis. Furthermore McKenzie therapy involves three basic steps: evaluation, treatment and prevention. LBP patients are divided in 3 subgroups: derangement, dysfunction and postural syndromes that are based on symptomatic and mechanical responses of the subjects to different movements and postures. Different exercises are involved in this therapy and they depend on the patient classification. The scope of the therapy is firstly to reduce and centralize the pain and then to eliminate completely the pain. After that the pain

has gone exercises for maintaining and preventing LBP are given to patients (McKenzie and May, 2003, Ponte et al., 1984);

- General exercising: this programme of exercising aims to improve general physical function, to decrease fear in using spine and to teach to patients how they should cope with the pain. This method included strengthening, stretching and aerobic exercises that involve all the main muscles of the body and not only back and abdominal muscles (Ferreira et al., 2007, Moffett and Frost, 2000).

2.10.3 Effectiveness of the different exercise programmes

Different exercise programmes have been proposed for LBP management, but it is not clear yet which programme is effective in reducing LBP symptoms (Airaksinen et al., 2006).

In the past one of the most prescribed cures for LBP was bed rest, but studies showed that staying active reduced time for recovery and risk of recurrence, decreasing also the social impact of this condition (Airaksinen et al., 2006, Malmivaara et al., 1995, Moffett et al., 1999, Waddell et al., 1997). It has been demonstrated that also mild aerobic activities and fitness programmes increased patient's outcome as decreased self reported pain and improved physical functioning, reducing need for physical therapy and medicine (Frost et al., 1998, Sculco et al., 2001). Furthermore, several studies demonstrated that fear avoidance beliefs can lead to avoidance in movement or activities, resulting in increased disability, in patients, (Asmundson et al., 1997, Grotle et al., 2004, Severeijns et al., 2001, Waddell et al., 1993) and, persistent symptoms in subjects with acute LBP (Burton et al., 1995, Fritz et al., 2001, Grotle et al., 2004).

Different types of exercises, described above, have been used in order to improve different aspects of the spine in LBP subjects (e.g. stability, mobility, muscular strength)

to reduce pain and disability due to this condition. Studies were performed to evaluate and compare different exercise programmes in order to test their effectiveness and to rationalise treatments.

After treatment outcomes, measured using self reported pain intensity and activity limitation, were compared between patients who followed a stabilisation exercising programme and patients who followed a general exercising programme and results showed that both programmes significantly reduce pain and disability in LBP patients but there were not different in the results between the two methodologies (Koumantakis et al., 2005, Unsgaard-Tøndel et al., 2010). Significant increased effectiveness (self reported questionnaire) was found for strengthening exercises when compared to patients that were instructed to remain active, but without being addressed a particular exercise programme (Hides et al., 2001b).

Trunk muscles strengthening exercises showed to improve significantly symptoms in LBP patients increasing muscle strength (Handa et al., 2000). Helewa, Goldsmith et al. (1999) evaluated the effects of strengthening exercises with back educational programme, comparing recurrence of LBP pain episodes in 2 years in patients that were divided in two groups: one that underwent both programme (strengthening and back educational) and another that underwent just back educational programme, finding that there were not differences in the results; Vasseljen and Fladmark (2010) compared strengthening exercises with general exercise using contraction thickness ratio for transversus abdominis, internal and external obliques muscles and self reported pain scores finding that both programmes improved patients' outcomes in the same manner. No differences in patients outcomes (mobility, pain and disability scores) were found when these exercises were compared with muscle coordination training (Johannsen et al., 1995).

Strengthening exercises were found to be more effective in reducing pain and disability in LBP when they were compared with passive or home exercising programme (Mayer et al., 2008).

Torstensen, Ljunggren et al (1998) showed that when LBP patients underwent mobilisation exercising programme, self reported outcomes (pain intensity, patient satisfaction, return to work, costs) were positive and were similar to LBP patients that used physiotherapy to treat their condition; furthermore this study showed that LBP patients who were just suggested to remain as active as before the pain without following neither the physiotherapy nor the mobilisation programmes outcomes were more negative in terms of time to recover and disability. No differences in the for LBP patients were also found in different types mobilisation exercises, in particular both flexion and extension exercises showed similar improvements in spinal mobility (Elnaggar et al., 1991) and in disability score and percentage of returning to work (Dettori et al., 1995).

A review article by Machado, de Souza et al. (2006) about McKenzie therapy showed several studies where this therapy was compared with strengthening (Petersen et al., 2002), mobilisation and general exercises programmes; results showed that patients reduced their pain intensity and improved their disability score in all the studies but there were no differences between patients treated with the McKenzie method and patients treated with the other methods. Contrasting results were found when McKenzie therapy was compared with Williams therapy: Ponte, Jensen et al.(1984) found better outcomes in LBP patients that underwent a McKenzie extension exercises when compared with patients treated with Williams flexion exercises, but these results were in contrast with the research of Dettori, Bullock et al. (1995) which found no differences in the outcomes of the two therapies.

Generalised exercise programme such as the “*Oxford fitness programme*”, which included different types of exercises (muscle strengthening, aerobic, stretching, mobilisation), showed to be more effective than traditional general practitioner management (Moffett et al., 1999) in reducing disability, pain intensity and use of healthcare services. Moreover, as mentioned above, general exercising was found to be as effective as exercise specific programmes or active physiotherapy (Helewa et al., 1999, Koumantakis et al., 2005, Machado et al., 2006, Mannion et al., 2001, Unsgaard-Tøndel et al., 2010, Vasseljen and Fladmark, 2010).

In summary exercising showed to be beneficial in reducing pain and in improving functionality, in particular mobility, muscle strength and endurance; furthermore exercise programmes showed to decrease back pain related disabilities also improving behavioural, cognitive and psychological factors and reducing concerns and fear related to the pain (Rainville et al., 2004).

Additionally, exercising was found to be more beneficial for LBP patients than bed rest or decreasing everyday activities, but specific exercises targeting specific aspects as mobilisation or stabilisation were not more effective than general exercising programmes.

It need to be highlighted that general exercising is not simply a suggestion to stay active or to perform self trained exercises, general exercise programmes are designed and performed under supervision of therapist (Moffett and Frost, 2000, Moffett et al., 1999) and they showed to more effective when prescribed after evaluation of patient characteristics (e.g. posture, muscle strength) and pain related symptoms (Descarreaux et al., 2002).

2.11 General summary

The review showed that alterations in different aspects of trunk biomechanics are present in LBP subjects when compared with asymptomatic subjects.

Studies evaluating trunk kinematics found alterations in subjects with LBP in variables such as joints ROM and velocity, joints contribution to the motion, joints coordination. These alterations were found to be dependent on the task, e.g. hip and lumbar spine ROMs were found decreased for basic trunk tasks such as bending but increased in walking or unstable sitting. Researchers explained these alterations as strategies in order to reduce pain and risks of further damages of the lumbar spine, but mechanism behind these alterations is not fully understood.

Postural control has been found altered in LBP patients and previous studies showed that postural control was poorer for LBP patients, who demonstrated increased postural sway and decreased percentage of successful task completion; furthermore when visual and vestibular system contributions to the control were removed or decreased and participants were relying mainly on the somatosensorial system, the postural control performance decreased more dramatically in LBP than in healthy participants, implying that LBP may affect trunk proprioception.

Viscoelastic properties of the trunk have been found to be altered in LBP patients. Mechanical characteristics of the discs have been evaluated and results showed that disc degeneration is involved in mechanical alteration of the tissues that may increase instability of the spine structures. Viscoelastic properties of the in-vivo trunk, especially stiffness coefficient, were found altered in LBP population. Increased trunk stiffness is believed to be a compensatory strategy to balance the decreasing in stiffness of the

spinal structures and as a strategy to reduce trunk motion to avoid risks of further damage and pain.

Increased muscular activity, altered temporal response and increased co-activation of the trunk muscles were found in LBP patients when compared with healthy participants. The reviewed studies showed and quantified alterations in symptomatic subjects but the mechanism behind these alterations has not been fully explained by the researchers.

Spinal loading and kinetics were altered in LBP subjects, in particular increase in spinal loading was found in LBP participants while performing different tasks (Kumar, 1990, Marras et al., 2001, Marras et al., 2004, Norman et al., 1998, Shum et al., 2007b). In-vitro studies showed that increasing in spinal stress may lead to damage and degeneration of the spinal structures (Adams et al., 1996, Gallagher et al., 2005). Furthermore kinematic and kinetic alterations such as decreased range of motion and power of the lumbar spine joint, were found between symptomatic and asymptomatic subjects and they were explained as compensatory strategy to reduce pain and decrease risks of dangerous movements (Marras et al., 2001, Marras et al., 2004, Shum et al., 2007b). Alterations were effective in decreasing pain, but there is not clear evidence that they are effective in decreasing tissues damage (Shum et al., 2007b); on the contrary, reduced mobility has been related to increased trunk muscles activation, that is well established to be related with increased spinal loading that may lead to injuries in spinal structures (Granata and Marras, 1995b, Marras et al., 2001, Marras et al., 2004).

Alterations in the posture have been found in LBP patients in sitting and standing posture. It is still unclear the causes/effects mechanism; Alterations in the

thoracolumbar curvature have been linked with different spinal disorders and LBP sufferers showed decreased lumbar lordosis. Further studies found relationship between working place and occurrence of LBP, especially in jobs where people had to sit for long period of time. The review also showed that exercise programme and postural correction need to be further investigate to evaluate if they are beneficial in improving LBP symptoms.

2.12 Limitations

LBP is one of the most wide spread pathological conditions and, as shown in the review, several studies have been performed in order to compare symptomatic with asymptomatic subjects. Limitations are present in these studies and results are contrasting, giving ambiguous conclusions, without given clear evidence to clinician. For example, there are contrasting interpretations about alterations in trunk muscular activity in symptomatic subjects: some studies gave the cause of these to the presence of muscles spasms or to the presence of neuromuscular impairments, other studies explain the increased activity as a protective strategy in order to stiffen the trunk and to reduce dangerous movement, but there is no clear evidence to support one of these theories.

Viscoelastic properties were limited to in-vitro studies for the evaluation of the mechanical properties of the spinal column and to in-vivo studies for passive properties (with no active muscles). Some studies were done evaluating these properties during active dynamic tasks, but these experiments used an external load attached to the participants or tasks that also involved lower limbs (as lifting a load from the floor) that could influence the results.

Alterations were also found in other biomechanical aspects of the trunk as kinematics, kinetics, postural control and posture variables but results were contrasting and related with the task.

2.13 Need for study

The literature review has highlighted the need to further investigate biomechanical trunk behaviour in non-specific LBP subjects, in particular dynamic properties of the spine, and how they are related to balance control and muscle activity. Evaluating and investigating the relationship between different biomechanical variables could help to understand the behaviour of the trunk and to give more explanation about how the differences in this behaviour affect different groups of subjects as healthy and LBP patients. As an example, evaluating muscular activity together with other biomechanical variables (e.g. kinematic, viscoelastic properties) could help in explaining the altered behaviour of the trunk and to clarify the role of the alteration in muscle activity in LBP subjects. On the other hand, investigating viscoelastic properties together with trunk kinetics, kinematics and muscle functioning will help to understand the mechanism behind alterations in trunk viscoelastic properties in LBP subjects.

Moreover, when performing an experiment that involves a precise task (e.g. regain balance after perturbation), the selection and performance of the task are important and could give more information about the effect of LBP on everyday life activities.

In addition the reviews highlighted that there is a lack of understanding regarding the action mechanism that may explain how exercise therapy improved the condition for chronic LBP. The relation between changes in clinical symptoms (e.g. self reported pain intensity or physical functioning) and functional characteristics of the trunk as range of

motion, muscular activity, dynamic properties (e.g. stiffness) has not yet been fully investigated.

Different studies showed that exercise programmes need to be assessed in order to evaluate if a certain programme is successful in challenging the target muscles without having dangerous effects (e.g. increasing spinal loading or angular deformation) to the patient (Axler and McGill, 1997, McGill, 1998, Scannell and McGill, 2003). As such, providing a methodology to measure the range of motion, muscular activity, spinal loading and dynamic properties of the trunk may be useful to evaluate, rationalise and improve treatments based on exercise therapy.

In order to investigate and evaluate effects of the LBP on the trunk biomechanical behaviour, overcoming the limitation of previous research shown in the literature review, a research design has been developed to meet the following requirements:

- Experimental task has to involve mainly trunk motion, to minimise the influence of other body parts (e.g. lower limbs);
- Viscoelastic properties of the trunk have to be evaluated while the participant is actively performing a task that mimics a common daily life activity with the trunk being the main part in conducting the task. This will give information about the overall mechanical properties of the trunk, whilst considering both passive and active structures.
- During the task, kinematics and kinetics of the trunk as well as trunk muscular activity need to be acquired at the same time;
- Performance of the experiment needs to be quantified using dedicated variables (e.g. length of the task, distance from the ideal target);
- Participants postural characteristics (e.g. lumbar lordosis, thoracic kyphosis) need to be acquired;

Results will serve to advance understanding of LBP and related biomechanical mechanisms in people with non-specific pain which, as a consequence, will enhance clinical work, helping clinician to rationalise and improve exercise and treatment for back pain increasing the benefits to the patients.

3 GENERAL METHODS

3.1 Introduction

This chapter provides a description of the design and building of the custom made swinging chair used for this research. The inverse dynamic equations used for the trunk kinematic and kinetic analysis are also presented. The last section describes the devices used for this research and their validation.

3.2 Custom made swinging chair

3.2.1 Design and building

The basic requirements for the design of the swinging chair were:

- Compatibility with the equipment used during the experiment;
- Reasonably lightweight for manual handling and to ensure low cost;
- Ability to be firmly mounted on the force platform
- High stability and rigidity;
- Ability to perform controlled swinging pattern;
- Swinging mechanism as frictionless as possible;
- Design adjustability for further corrections and additions throughout the project;
- Suitable for wide range of participant weights and heights;
- Foot and leg support and security belt to immobilize the legs and pelvis;

According to these requirements, plywood was chosen as the main material to build the apparatus. The base of the chair was designed in order to meet the dimension of the force platform because it needed to rest completely on it, without touching the floor. Four holes were drilled on the base in order to secure the base onto the place using fixing screws. The swinging mechanism was provided by two low friction ball bearings. Mechanical stops prevent the chair from swinging more than 20 degrees backward and

forward. The seat had adjustable belts and footrest in order to restrict movements of the trunk downwards and to prevent the risks of falling. The seat was designed with a sliding mechanism to allow the participant to balance in the resting position. A handle was placed on the chair in order to tilt the chair backward and forward. The chair was built with two main parts:

- Static part: formed by the base (that permitted attachment with the force plate) and two vertical supports attached to the base, where the two bearings were placed to connect the swinging seat);
- Swinging part: formed by the seat and by the leg and foot supports. This part was consisted of two cylindrical pieces of wood that were fitted into the bearing holes (the hinges) to permit swinging.

The two parts of the chair were built in a workshop, then assembled and modified in the laboratory. Figure 3-1 and Figure 3-2 show the custom made swinging chair.

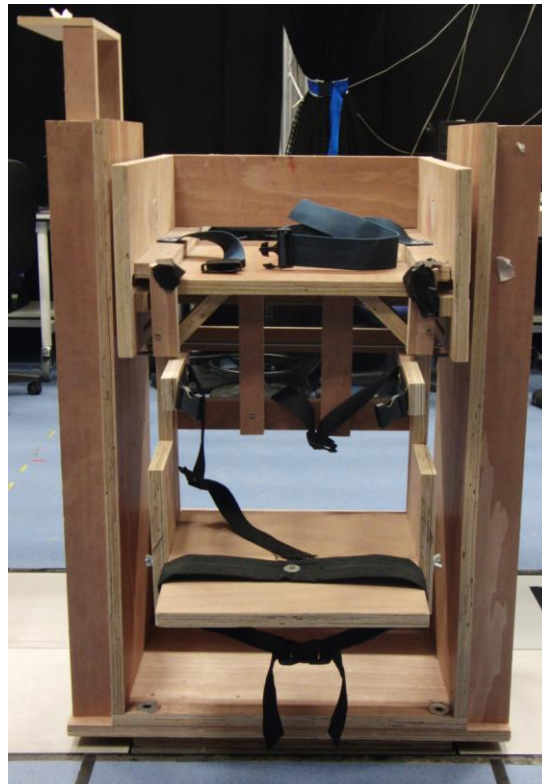


Figure 3-1: Front view of the custom made chair

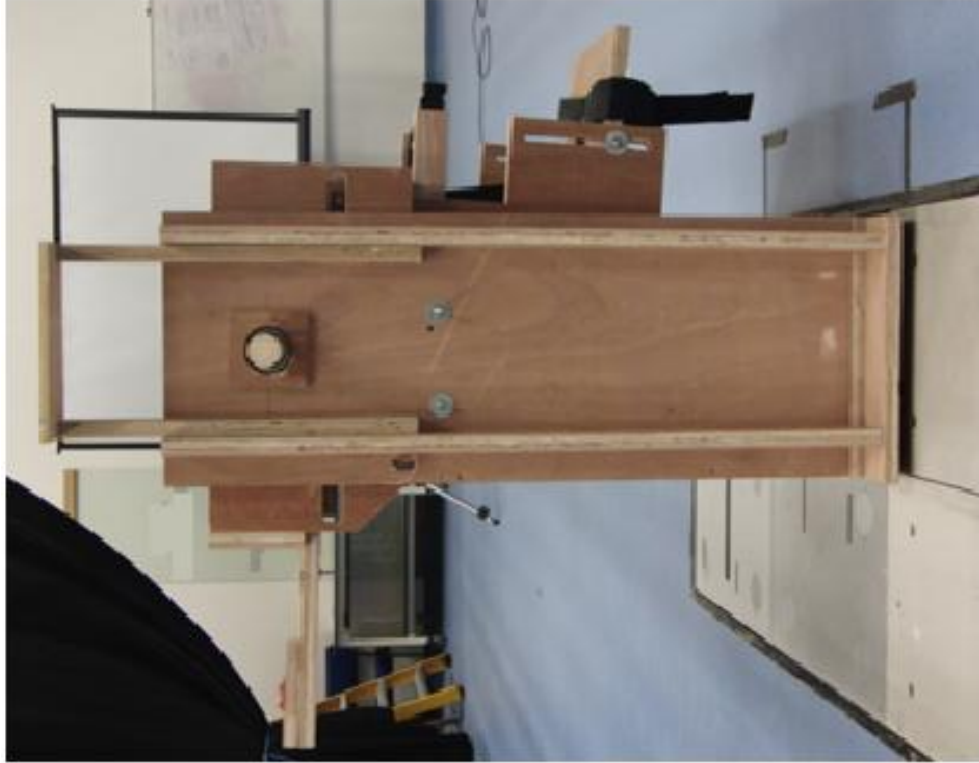


Figure 3-2: Rear and side view of the custom made chair

3.2.2 Calculation of the weight, centre of mass and inertia

The mechanical characteristics of the chair were required for the inverse dynamic analysis. The total weight of the swinging chair, the weight of the static part, and the weight of the swinging part were required. The weight of the support part of the chair was measured on the force platform before chair assembly. The chair was assembled and placed on the force platform and the total weight of the chair was measured. Values of the weight of the chair are shown in table C1. In addition, the coordinates of the centre of mass of the swinging part of the chair in static position (with the seat parallel to the floor) were also required for the data analysis. As for the weight, the centre of mass of the swinging part of the chair was derived from the total chair and support part centre of mass. The X and Y coordinate of the centre of mass of the support part and of the total chair in the force platform coordinate system were measured using the force platform software. The Z coordinate of the centre of mass was not provided by the software and so it was calculated through experimental data. The support part was slightly tilted at a known angle over the force platform and the X and Y coordinates of the centre of mass of this part in this position was measured. Using the X and Y coordinates of the centre of mass when the support was not tilted and when it was tilted about 9 degrees the Z coordinate was calculated by:

Equation 3-1

$$Z_{Scm} = \frac{(Y_{Scm} - Y_{Scm9})}{\tan 9}$$

Where:

Z_{Scm} , Y_{Scm} = Y and Z coordinates of the support part in static position centre of mass;

Y_{Scm9} = Y coordinate of the support part tilted about 9 degrees centre of mass;

The X, Y and Z coordinates of the centre of mass of the swinging chair was calculated as follows:

Equation 3-2

$$X_{SWcm} = \frac{W_T * X_{Tcm} - W_{Scm} * X_{Scm}}{W_T}$$

Equation 3-3

$$Y_{SWcm} = \frac{W_T * Y_{Tcm} - W_{Scm} * Y_{Scm}}{W_T}$$

Equation 3-4

$$Z_{SWcm} = \frac{W_T * Z_{Tcm} - W_{Scm} * Z_{Scm}}{W_T}$$

Where:

$X_{SWcm}, Y_{SWcm}, Z_{Wcm}$ = X, Y and Z coordinates of the swinging part in static position centre of mass;

X_{Scm} = X coordinate of the centre of mass of the support part in static position;

$X_{Tcm}, Y_{Tcm}, Z_{Tcm}$ = X, Y and Z coordinates of the total chair in static position CM;

W_T = total weight of the chair; W_S = weight of support part;

W_{SW} = weight of the swinging part;

Table 3-1: Weight and CM coordinates in the force platform coordinate system for swinging part, static part and total chair

	Swinging part	Static part	Total
X-coordinate CM (m)	0.0088	0.0201	0.0160
Y-coordinate CM (m)	-0.0062	-0.0047	-0.0053
Z-coordinate CM (m)	0.7519	0.3594	0.5032
Weight (N)	182.76	316.16	489.92

The moment of inertia in the sagittal plane of the swinging part of the chair was required for data analysis. The value was calculated through the pendulum test. This method has been already used to calculate the moment of inertia other objects (Fowles and

Cassidy, 2004). An accelerometer was mounted on the backrest of the chair, in order to measure the oscillation during the experiment. The chair was tilted forward and released. Angular displacement was recorded until the chair stopped swinging.

Using the equation of the period of oscillation of physical pendulum inertia of the swinging chair about the axis of rotation (the hinges of the chair) in the sagittal plane was calculated as follow:

Equation 3-5

$$I = \frac{T^2 * M_s * g * d}{4 * \pi^2}$$

Where

I = inertia about the rotational axes; T = Period of oscillation;

Ms = Mass of the swinging part of the chair; g = gravity;

d = distance between rotational axes and the swinging part of the chair centre of mass;

Ms, g, d were constants so the equation became: $I = T^2 * 1.283$;

The distance d was calculated with the coordinates of centre of mass of the swinging part of the chair and the coordinates of the centre of rotation in the force platform coordinate system. In order to check the validity and the reliability of this method the inertia has been calculated with different tilting angles (considering the resting point as 0 degrees, the angles used were 1.9, 2.4, 4, 5, 7, 8, 10, and 20) and the calculation was completed using the first, second and third period of oscillation during experiment. In Figure 3-3 the oscillation angle for one of the test used is shown; the periods of oscillation marked as T1, T2 and T3 and their values are reported in seconds. Values of the moment inertia calculated with each time period are also shown in the figure. The mean of the moment of inertia of all experiments was 3.65 kg m² (SD ±0.022) and this value was used in the inverse dynamic equations.

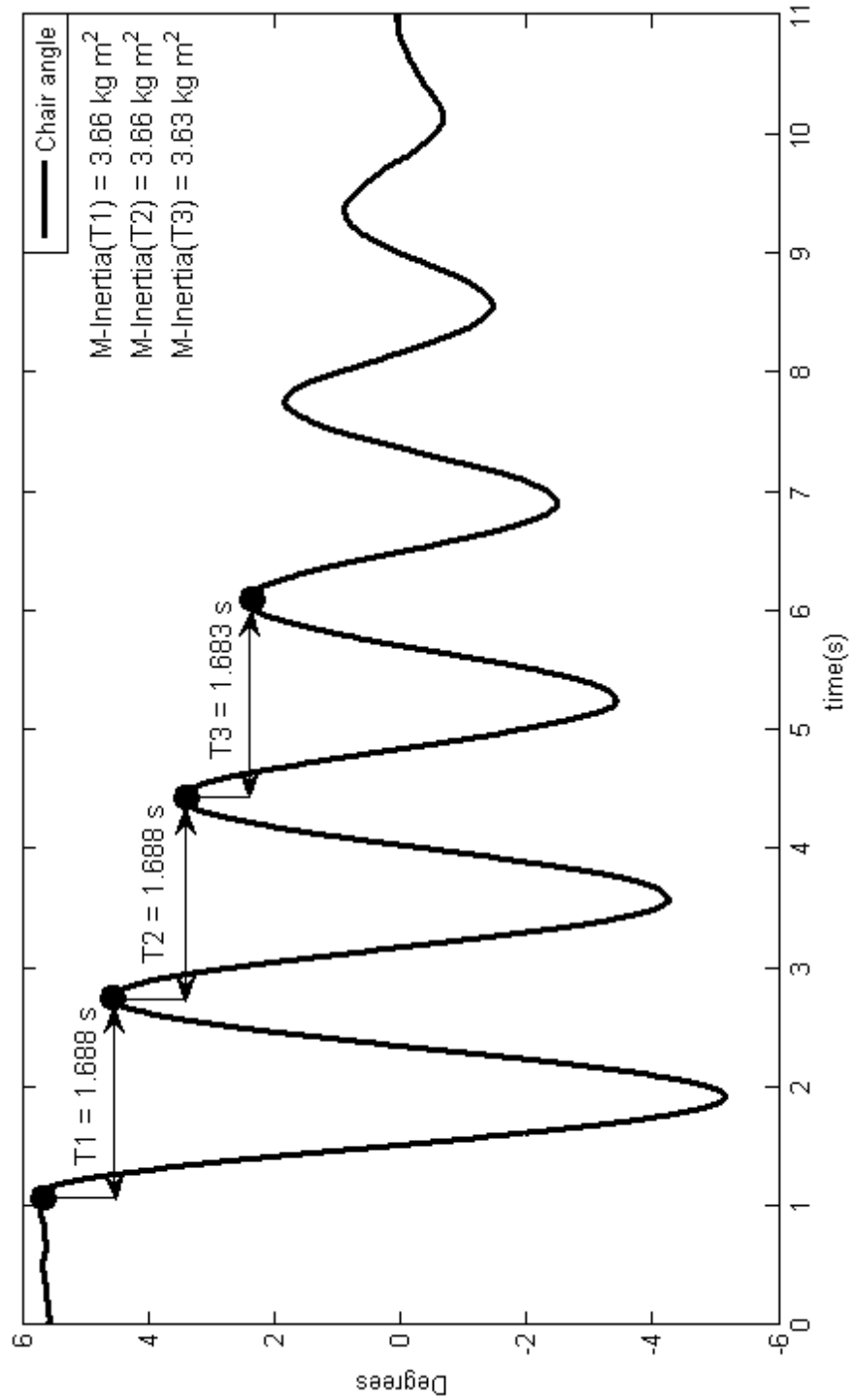


Figure 3-3: Oscillation angle for the moment of inertia calculation; T1, T2 and T3 are the oscillations periods and M-Inertia(T1), M-Inertia(T2) and M-Inertia(T3) the moment of inertia calculated for this trial using the three oscillations periods.

3.3 Inverse dynamic analysis

The data collected from the motion tracking system and the force platform were combined to obtain the value of the moment acting at hip and lumbar spine joints level.

3.3.1 Re-sampling of the data

Dynamic data were re-sampled because the motion tracking system returned uneven data and the force platform data were acquired at 150 Hz sampling frequency. All data were re-sampled at 40 Hz with Matlab software using the spline function.

3.3.2 Coordinate system

Firstly, in order to combine the data, the same coordinate system needed to be the same for both the devices. The coordinates system was defined as (considering the position of the body on the chair) as follows:

Z= vertical direction (positive from the roof to the floor);

Y= anterior-posterior direction (positive direction from posterior to anterior);

X= perpendicular to the Z direction and pointing right;

The origin of the coordinate system was the fastrak transmitter.

3.3.3 Inverse dynamic analysis

3.3.3.1 Hip reference point static (point H)

The moment was calculated in the caudal endpoint of the trunk, which according with Zatsiorsky (2002) was defined as *“the intersection of the projections on the frontal plane of the hip segmentation plane”* which were, in turn, defined as *“the boundaries between thighs and trunk, defined as planes passing through the respective iliospinales, parallel to the trunk sagittal axis, and forming a 37 degrees angle with the sagittal plane”*.

The coordinates of this point (called H) were calculated using the digitised left and right ASISs and PSISs. The X coordinate was calculated as the mean between the midpoint between left and right ASISs and the midpoint between left and right PSISs. The Y and Z coordinates calculation were calculated twice: firstly using right ASIS and PSIS and then using left ASIS and PSIS. In order to get the x and z coordinates the mean of the left and right side was calculated. The formulae below show how the Z and Y coordinates were calculated for the right side. The same formulas were used for the left side.

Equation 3-6

$$c1 = (RPSIS_z - RASIS_{xz});$$

Equation 3-7

$$c2 = \frac{c1}{\tan 53^\circ};$$

Equation 3-8

$$K = RASIS_y - c2;$$

Equation 3-9

$$Y_{Hr} = \frac{K + RPSIS_y}{2};$$

Equation 3-10

$$c3 = Y_{Hr} - RPSIS_y;$$

Equation 3-11

$$c4 = c3 * \tan 53^\circ;$$

Equation 3-12

$$Z_{Hr} = RPSIS_z + c4;$$

Where

$RPSIS_z$, $RPSIS_y$ = z and y coordinates of right PSIS;

$RASIS_z$, $RASIS_y = z$ and y coordinates of right ASIS;

Y_{Hr} , $Z_{Hr} = Y$ and coordinates of the right point H;

K , c_1 , c_2 , c_3 and c_4 are represented in Figure 3-4.

Z and Y coordinates of the H point were calculated as the midpoint between the H point calculated for the right side and the H point calculated for the left side using the formulas above.

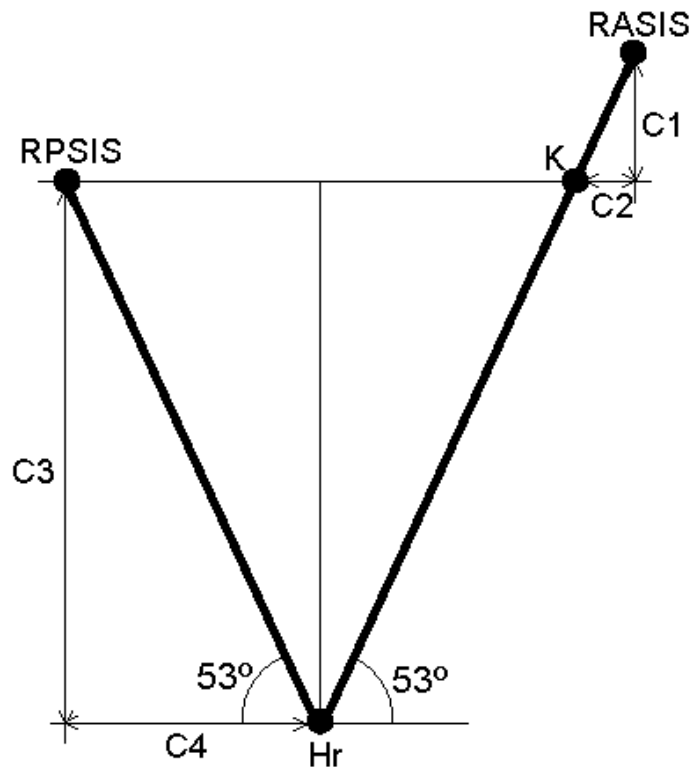


Figure 3-4: Hip reference point static calculation, example for the right side

3.3.3.2 L5-S1 joint centre static

The position of the L5-S1 centre joint was estimated using the digitised ASISs and PSISs. The L5-S1 joint centre was defined as a point lying 40% of the distance (from posterior to anterior) between the two midpoints of the two posterior superior iliac spines (PSIS) and the midpoint of the anterior superior iliac spines (ASIS) (Gallagher et

al., 2009). Following this definition the coordinates of the L5-S1 joint centre were calculated in the following manner:

Firstly the midpoint between the left and right ASISs in all three coordinates (x, y and z) were calculated using the digitised LASIS and RASIS coordinates:

Equation 3-13

$$\text{midA}_x = \frac{\text{LASIS}_x - \text{RASIS}_x}{2} + \text{RASIS}_x$$

Equation 3-14

$$\text{midA}_y = \frac{\text{LASIS}_y - \text{RASIS}_y}{2} + \text{RASIS}_y$$

Equation 3-15

$$\text{midA}_z = \frac{\text{LASIS}_z - \text{RASIS}_z}{2} + \text{RASIS}_z$$

Then the midpoint between the left and right PSISs in all three coordinates (x, y and z) were calculated using the digitised LPSIS and RPSIS coordinates:

Equation 3-16

$$\text{midP}_x = \frac{\text{LPSIS}_x - \text{RPSIS}_x}{2} + \text{RPSIS}_x$$

Equation 3-17

$$\text{midP}_y = \frac{\text{LPSIS}_y - \text{RPSIS}_y}{2} + \text{RPSIS}_y$$

Equation 3-18

$$\text{midP}_z = \frac{\text{LPSIS}_z - \text{RPSIS}_z}{2} + \text{RPSIS}_z$$

The x, y and z coordinates of the L5-S1 joint centre were then calculated as follows:

Equation 3-19

$$\text{L5S1}_x = \text{midP}_x + \frac{\text{midP}_x - \text{midA}_x}{2}$$

Equation 3-20

$$L5S1_y = \text{midP}_y + \frac{\text{midP}_y - \text{midA}_y}{2.5}$$

Equation 3-21

$$L5S1_z = \text{midP}_z + \frac{\text{midP}_z - \text{midA}_z}{2}$$

3.3.3.3 Centre of mass static

The location of the position of the centre of mass of the feet, shanks, thighs and pelvis were found as discussed by Zatsiorsky (2002), using digilised points and anthropometric data.

In order to calculate the centre of mass, the participant sat on the chair in a fixed position with the angle between foot and shank, the shank and the thigh, and the thighs and trunk at about 90 degrees as shown in Figure 3-5. The coordinates of the centre of mass of the right thigh were calculated with the following equations:

Equation 3-22

$$\text{Dist}_{\text{icm-tr}_y} = \text{tlr} * 1.083 * P_t$$

Equation 3-23

$$\text{Cm}_{\text{tr}_y} = \text{rasis}_y + \text{Dist}_{\text{icm-tr}_y}$$

Equation 3-24

$$\text{Cm}_{\text{tr}_z} = \text{tror}_z;$$

Where:

$\text{Dist}_{\text{icm-tr}_y}$ = distance between centre of mass and anterior superior iliac spine right in the y coordinate;

tlr = length of the right thigh;

$tror_z$ = z coordinate of the right trochanterion;

P_t = coefficient for location of centre of mass of the thigh, different for male and female;

Cm_{tr_z} , Cm_{tr_y} = z and y coordinates of the centre of mass of the right thigh.

The coordinates of the centre of mass of the right shank were calculated with the following equations:

Equation 3-25

$$Cm_{sr_y} = rasis_y + (tlr * 1.083)$$

Equation 3-26

$$Cm_{sr_z} = tror_z + (srl * P_s)$$

Where:

slr = length of the right shank;

P_s = coefficient for location of cm of the shank, different for male and female;

Cm_{sr_z} , Cm_{sr_y} = z and y coordinates of the centre of mass of the right shank.

The coordinates of the centre of mass of the right foot were calculated with the following equations:

Equation 3-27

$$Cm_{fr_y} = (frl * P_f) - har + rasis_y + (tlr * 1.083)$$

Equation 3-28

$$Cm_{fr_z} = tror_z + srl$$

Where:

fll = length of the right foot; har = distance between heel and the sphyron fibulare;

P_f = coefficient for location of cm of the foot, different for male and female;

Cm_{fr_z} , Cm_{fr_y} = z and y coordinates of the centre of mass of the right foot.

The same calculation was done for the left side in order to get the centre of mass coordinates of the left thigh, shank and foot. The coordinates of the centre of mass of the pelvis were calculated with the following equations:

Equation 3-29

$$Cm_{pz} = (H_z - (pl * 2.305)) + (2.305 * pl * 0.3541)$$

Where:

pl = length of the pelvis;

Cm_{pz} = z coordinate of the centre of mass of the pelvis.

$Cm_{py} = Y_{Hr}$ = coordinate of the centre of mass of the pelvis.

Lengths of the body part and coefficients were taken from Zatsiorsky (2002).

The coordinates of the total centre of mass of lower limbs and the swinging part of the seat were calculated with the following formulas:

Equation 3-30

$$Cm_{lowz} = \frac{\sum m_i * CMZ_i}{\sum m_i}$$

Equation 3-31

$$Cm_{lowy} = \frac{\sum m_i * CMY_i}{\sum m_i}$$

Where:

i represents in turn left and right foot, shank and thigh and the swinging part of the seat;

m_i is the weight of a certain i element;

CMZ_i and CMY_i are respectively the z and y coordinates of the centre of mass of the i element. The weight of each body part was calculated using the following formula:

Equation 3-32

$$m_j = km * L_j * C_j^2$$

Where:

j = it represents in turn left and right foot, shank and thigh and the pelvis;

k_m = coefficient for the mass calculation, different for male and female participants and for different body parts;

L_j = length of a certain body parts; C_j = circumference of a certain body parts.

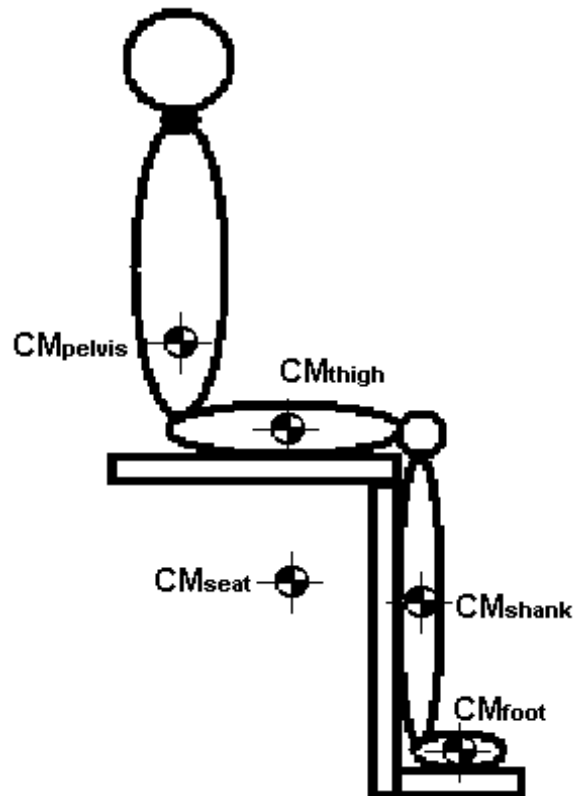


Figure 3-5: Centre of masses location in static position

3.3.3.4 Moment inertia of the body segments

Firstly the inertia of each body part respect to their centre of mass was calculated with the following formula:

Equation 3-33

$$I_j = k_s * m_j * L_j^2$$

I_j = inertia of a certain body parts respects its CM;

k_s = coefficient for the inertia calculation from Zatsiorsky (2002), different for male and female participants and for different body parts.

The inertia used for the calculation of the moment was calculated in respect to the point H (hip reference point). The general equation to calculate the inertia in respect to the point H was as follows:

Equation 3-34

$$d_{cm-H} = \sqrt{(H_z - Cm_{j_z})^2 + (H_y - Cm_{j_y})^2}$$

Equation 3-35

$$I_{H_j} = I_j - (m_j * d_{cm-H}^2)$$

Where:

d_{cm-H} is the distance between the point H and centre of mass of a certain body part j;

I_{H_j} = inertia of a certain body part j in respect to the point H.

The same formulas were used to find the inertia of the swinging part of the seat in respect to the point H. The inertia of the pelvis was calculated in respect to the L5-S1 joint centre as follow:

Equation 3-36

$$d_{cmPel-L5S1} = \sqrt{(Cm_{pz} - L5S1_z)^2 + (Cm_{py} - L5S1_y)^2}$$

Equation 3-37

$$I_{pL5S1} = I_p - (m_p * d_{cmPel-L5S1}^2)$$

Where:

$d_{cmPel-L5S1}$ is the distance between the L5-S1 joint centre and centre of mass of the pelvis;

I_{pL5S1} = inertia of the pelvis in respect to L5-S1 joint centre.

3.3.3.5 Seat and pelvis angular acceleration - Cm_{low} linear acceleration

In order to calculate linear and angular acceleration, a 5 point differentiation method was used. The formula for the 5 point differentiation method was as follow:

Equation 3-38

$$f'' = \frac{-f(n+2) + 16 * f(n+1) - 30 * f(n) + 16 * f(n-1) - f(n-2)}{12 * t^2}$$

Where

f = it is the function whose second derivate needed to be calculated;

t = it is the inverse of the sample frequency (in this experiment the frequency is 40 Hz and so t is 0,025 s).

This formula was used to calculate:

- Seat angular acceleration (ε_s) with f = angular displacement in the sagittal plane coming from the sensor placed on the swinging part of the chair;
- Pelvis angular acceleration (ε_p) with f = angular displacement in the sagittal plane coming from the sensor placed on the first sacral vertebrae;
- Y component of the linear acceleration of the Cm_{low} (Acc_{cm_{low}y}) with f = y component of the linear displacement of the Cm_{low} during the trial;
- Z component of the linear acceleration of the Cm_{low} (Acc_{cm_{low}z}) with f = z component of the linear displacement of the Cm_{low} during the trial.

3.3.3.6 Hip reference point dynamic (point H)

The coordinates of the point H calculated previously for the static position were combined with the data from the sensor on the seat in order to have the coordinates of

this point during the trial. The angular displacement in the sagittal plane coming from the sensor placed on the swinging part of the chair was used.

The distance between centre of rotation (the hinge of the custom made chair) and the static point H was calculated:

Equation 3-39

$$\text{distH} = \sqrt{(R_z - H_z)^2 + (R_y - H_y)^2}$$

Then the x and z coordinates of the centre of mass were calculated as follows:

Equation 3-40

$$\text{DynH}_z = R_z + (\text{distH} * \sin(\theta_s - \theta_{\text{fixed}}))$$

Equation 3-41

$$\text{DynH}_y = R_y + (\text{distH} * \cos(\theta_s - \theta_{\text{fixed}}))$$

Where:

$\theta_{\text{fixed}} = \cos^{-1} \frac{(R_z - H_z)}{\text{distH}}$; R_z , R_y = z and y coordinates of the centre of rotation;

θ_s = angular displacement in the sagittal plane from the sensor placed on the swinging part of the chair;

DynH_z , DynH_y = z and y coordinates of the centre of mass during the trial.

3.3.3.7 Centre of mass dynamic condition

The coordinates of the total centre of mass of lower limbs plus swinging part of the seat calculated previously for the static position were combined with the data from the sensor on the seat in order to have the coordinates of this centre of mass during the trial in the same manner as for the point H (Hip reference point static).

The distance between centre of rotation (the hinge of the custom made chair) and the static centre of mass was calculated:

Equation 3-42

$$\text{dist} = \sqrt{(R_z - \text{Cm}_{\text{low}_z})^2 + (R_y - \text{Cm}_{\text{low}_y})^2}$$

Then the x and z coordinates of the centre of mass were calculated as follows:

Equation 3-43

$$\text{DynCm}_{\text{low}_y} = R_y - (\text{dist} * \cos(\theta_s - \theta_{\text{fixed}}))$$

Equation 3-44

$$\text{DynCm}_{\text{low}_z} = R_z + (\text{dist} * \sin(\theta_s - \theta_{\text{fixed}}))$$

Where:

$\theta_{\text{fixed}} = \cos^{-1} \frac{(R_z - \text{Cm}_{\text{low}_z})}{\text{dist}}$; $\text{DynCm}_{\text{low}_y}$, $\text{DynCm}_{\text{low}_z}$ = y and z coordinates of the centre of mass during the trial.

3.3.3.8 L5-S1 joint centre dynamic

The position of the L5-S1 joint centre during the trial was derived combining the static position of the joint and the data from the sensor in the 1st sacral vertebrae. The distance between the first sacral vertebrae and the L5-S1 joint centre was assumed to be fixed throughout the trial. As a result, the motion of the joint centre was related to the motion of the sensor. A static measure of the sensor in the sacrum was taken before starting the trial. The data about the orientation (azimuth, elevation and roll angles) and position (x, y and z coordinates) of the sensor in the sacrum for the static position were used to transfer the data of the static L5-S1 joint centre from the global coordinate

system to a local coordinate system (with the origin centred in the position of the sensor in the sacrum) as follow:

Equation 3-45

$$[R] * \begin{bmatrix} L5S1_x \\ L5S1_y \\ L5S1_z \end{bmatrix} + \begin{bmatrix} S1_x \\ S1_y \\ S1_z \end{bmatrix} = \begin{bmatrix} L5S1_{xl} \\ L5S1_{yl} \\ L5S1_{zl} \end{bmatrix}$$

Where:

$[R]$ = rotation matrix made with data about the orientation of the sensor in the sacrum for the static position;

$\begin{bmatrix} S1_x \\ S1_y \\ S1_z \end{bmatrix}$ = coordinates of the sensor in the sacrum for the static position;

$\begin{bmatrix} L5S1_{xl} \\ L5S1_{yl} \\ L5S1_{zl} \end{bmatrix}$ = coordinates of the joint centre in the local coordinate system.

Translation parameters and rotation matrix for each frame of the trial were calculated using the dynamic data of the sensor on the sacrum. The matrix and the parameters were combined with the L5-S1 joint centre position in the local coordinate system in order to get the position of the joint throughout the trial in the global coordinate system.

The equation for each frame was as follows:

Equation 3-46

$$\begin{bmatrix} [R_f]^t * \begin{bmatrix} L5S1_{xl} \\ L5S1_{yl} \\ L5S1_{zl} \end{bmatrix} \end{bmatrix} + \begin{bmatrix} S1_{dyn_x} \\ S1_{dyn_y} \\ S1_{dyn_z} \end{bmatrix} = \begin{bmatrix} L5S1_{dyn_x} \\ L5S1_{dyn_y} \\ L5S1_{dyn_z} \end{bmatrix}$$

Where:

$[R_f]^t$ = transpose of the rotation matrix made with data about the orientation of the sensor in the sacrum from one single frame of the trial;

$$\begin{bmatrix} S1dyn_x \\ S1dyn_y \\ S1dyn_z \end{bmatrix} = \text{coordinates of the sensor in the sacrum from one frame;}$$

$$\begin{bmatrix} L5S1dyn_x \\ L5S1dyn_y \\ L5S1dyn_z \end{bmatrix} = \text{coordinates of the joint centre in global coordinate system from one frame.}$$

The calculation was done for each frame in order to get the data of the position of the L5-S1 joint centre throughout the trial.

3.3.3.9 Dynamic centre of mass of the pelvis

The position of the centre of mass of the pelvis during the trial was calculated using the same equations that were used for the position of the L5-S1 joint centre throughout the trial. Also in this calculation the data from the sensor on the sacrum were used. The static L5-S1 position was substituted with the static position of the centre of mass of the pelvis.

3.3.3.10 Moment in the point H (hip moment)

The moment acting in the caudal endpoint of the trunk (point H) was calculated using inverse dynamic equations. The forces measured from the force platform were assumed to be applied on the centre of rotation of the seat (the hinge).

In the Figure 3-6 the free body diagram is shown for the calculation of the hip moment.

According with the Figure 3-6 the equation used was:

Equation 3-47

$$M_H = I_H * \varepsilon_S - F_Z * d_1 + F_y * d_2 + m_L * g * d_3$$

Where:

$d_1 = R_y - H_y$; $d_2 = R_z - H_z$; $d_3 = C m_{low_y} - H_y$; ε_S = angular acceleration of the seat;

I_H = total inertia at the point H calculated as the sum of the inertia of left and right foot, shank, thigh and swinging part of the chair;

F_y , F_z = y and z components of the force during the trial;

m_L = sum of the mass of left and right foot, shank, thigh and swinging part of the chair;

g = gravitational acceleration; M_H =moment in the point H.

The moment was calculated for each frame throughout the trial.

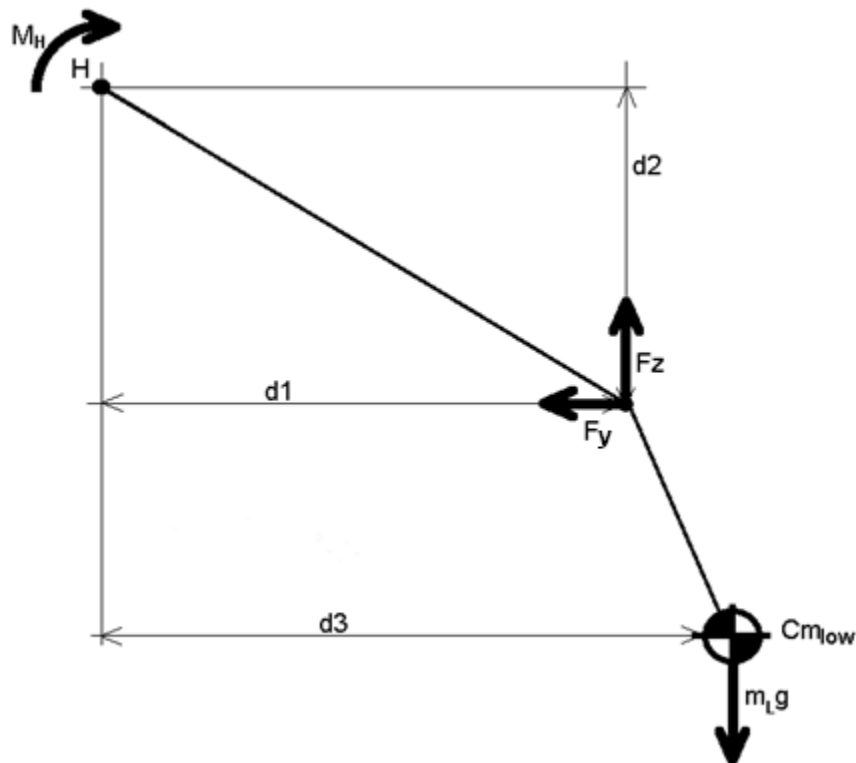


Figure 3-6: Free body diagram for the calculation of the moment in the Hip

3.3.3.11 Moment in the L5-S1 joint centre (lumbar spine joint moment)

The moment acting in the L5-S1 joint centre was calculated using inverse dynamic equations. From the Figure 3-7 forces acting in the point H were calculated as follows:

Equation 3-48

$$m_L * a_{CM_y} + F_y = F_{H_y}$$

Equation 3-49

$$m_L * (a_{CM_z} - g) + F_z = Fh_z$$

Where:

a_{CM_z}, a_{CM_y} = z and y components of the linear acceleration of the centre of mass of the lower limbs and the swinging part of the seat;

In the Figure 3-7 is shown the free body diagram for the calculation of the moment.

According with the Figure 3-7 the equation used was:

Equation 3-50

$$M_{L5S1} = I_P * \varepsilon_P + M_R + Fr_z * d_4 - Fr_y * d_5 + m_p * g * d_6$$

Where:

$$d_4 = H_y - L5S1_y; d_5 = H_z - L5S1_z; d_6 = C_{m_p} - L5S1_y;$$

ε_P = angular acceleration of the pelvis; I_P = inertia of the pelvis; $Fr_y = Fh_y$; $Fr_z = Fh_z$;

m_p = mass of the pelvis; $M_R = M_H$; M_{L5S1} = moment in L5S1 point.

The moment was calculated for each frame throughout the trial.

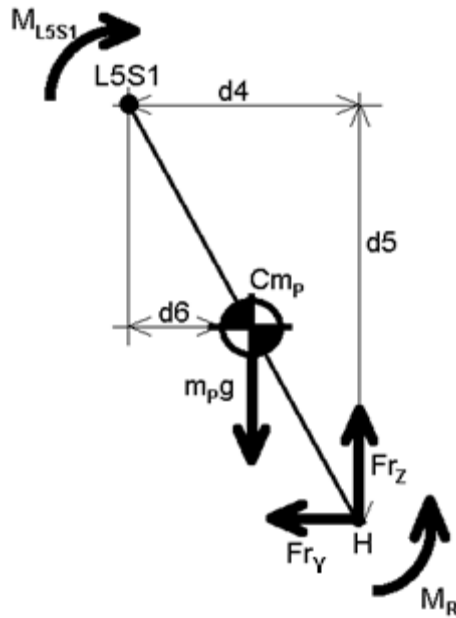


Figure 3-7: Free body diagram for the calculation of the moment in L5S1 joint centre

3.3.4 MatLab coding

The formulas presented in the inverse dynamic equations section above were inserted in a custom made MATLAB® (R2007b, MathWorks Inc.) code where data from the force platform and motion tracking system sensors were inserted to calculate hip and lumbar spine kinematic and kinetic. The code permitted to maximise data elaboration time and to standardise data analysis.

3.4 Equipments and their validation

3.4.1 Electromagnetic motion track system

3.4.1.1 Features

The movements of the lumbar spine were measured using a three-dimensional motion track system (3SPACE FASTRAK®, Polhemus Inc.). Electromagnetic fields are used to determine the position and orientation of a remote object. This is achieved by a fixed transmitter generating low frequency magnetic field vectors from an assembly of three concentric stationary antennas. The field vectors are detected by a mobile sensing antenna i.e. a receiver. The signals detected are then processed by a mathematical algorithm that returns the position and orientation of the receivers relative to the transmitter. The apparatus consists of a System Electronics Unit (SEU), the transmitter, four receivers and the power supply. The Transmitter and receivers cannot be directly in contact with metal surfaces because metal could interfere with the magnetic field. The FASTRAK system is connected to a computer via serial port (RS-232) or USB serial communication. The operating frequency varies according to the number of receivers that are simultaneously used i.e. with one receiver the system works at

120Hz, with two at 60 Hz, with three at 40 Hz and with four receivers at 30 Hz. The accuracy was checked in validation tests that will be described later in this chapter. Data were extracted using ASCII files. Inches were used as unit for linear displacement and degrees were used as unit for angular displacement.

3.4.1.2 Validation test

Previous studies, described in chapter 2, showed that the characteristics of the electromagnetic motion system are dependent on the environment and they can be altered from the one reported on the instruction manual. The reliability was checked for the electromagnetic motion track system in order to evaluate the compatibility with the experimental environment. Two motion sensors were mounted on a piece of wood at a fixed distance (313.2 mm) and then placed on the swinging chair. The chair was placed over the force platform and the test was conducted with the same set-up and protocol of the main experiment. The test was repeated by placing the piece of wood in different locations over the seat. The chair was tilted backward and forward about 20 degrees and data of the motion of the two sensors were recorded. The trial was repeated three times. The distance between the two sensors was calculated as the square root of the sum of the difference of the position of each coordinate with the following equation:

Equation 3-51

$$d = \sqrt{(X1 - X2)^2 + (Y1 - Y2)^2 + (Z1 - Z2)^2}$$

Where X1, Y1 and Z1 are the position coordinate of sensor 1 and X2, Y2 and Z2 were the position coordinates of sensor 2. Mean and standard deviation of the error was calculated as the differences between known fixed distance and distance calculated during motion. In Table 3-2 mean of the measured distance value, mean of the error as percentage of the known length and maximum error during the test as percentage of the known length are shown. The mean of the error across the 5 tests was 0.54 % of

the known distance with a maximum error of 3.41%. This validation test indicated the reliability of the motion system and that the area used for the experiment was free from electromagnetic interference.

Table 3-2: Mean of the length value and error % and maximum error percentage for the 5 validation tests

	Test 1	Test 2	Test 3	Test 4	Test 5
Mean value (mm)	311.4	311.6	311.6	310.6	312.4
Mean error (%)	0.57	0.51	0.51	0.83	0.26
Max error (%)	2.40	2.68	3.41	1.79	2.90

3.4.2 Surface EMG

3.4.2.1 Features

The system used to detect the electrical activity of the muscles consists of:

- Data acquisition system DataLINK (Biometrics Ltd type Nos. LS800);
- 8 precision bipolar and differential EMG surface electrodes (Biometrics Ltd type Nos. SX230);
- Subject unit to connect the acquisition system with the sensors;
- Biometrics Analysis software

The subject unit can acquire data from 8 EMG sensors; it is connected to a computer through the DataLINK acquisition system. The subject unit is programmable via dedicated software (Biometrics Analysis software) that allows to the user to select the gain, sampling frequency, sensor supply voltage, and zero or threshold level and hysteresis level for each channel individually. The analogue outputs from the basic unit can be connected to any analogue recording system through an output cable (Biometrics R2000i). The SX230 sensor consists in two integral, dry, reusable, circular

electrodes (with inter electrode distance of 20 mm and diameter of 10 mm) connected to a differential amplifier with 1000 gain, and a filter that limit the bandwidth from 20Hz to 450 Hz. The actual gain of the sensors is 1000, the common mode rejection ratio (CMRR) > 96 dB and its impedance is very high (Input Impedance of $> 10^{15}$ Ohms), and so use of conducting gels can be avoided and need of skin preparation minimised. A single Ag/AgCl disposable electrode was used as reference.

3.4.2.2 Interference between EMG and motion track system tests

Validation tests were completed to check possible unwanted interactions between EMG and electromagnetic system. The same experiment used above was replicated with the EMG sensors positioned next to the motion sensors in order to check possible malfunctioning in the motion tracking device. The chair was tilted and released and distance between the sensors measured in the same manner as it is reported in the section above. Data acquired with and without the EMG turned on were compared in order to check possible interference. Differences found in motion data were less than 0.5 mm in the 3 tests performed as it is shown in Table 3-3.

Table 3-3: Results of the EMG interference in motion tracking device data tests

	Test 1		Test 2		Test 3	
	EMG ON	EMG OFF	EMG ON	EMG OFF	EMG ON	EMG OFF
Distance (mm)	311.4	311.7	311.5	311.4	312.5	312.3
Mean \pm SD	± 1.3	± 2.0	± 2.7	± 3.0	± 2.6	± 2.3
Difference (mm)	0.3		0.1		0.2	

Another experiment was performed in order to check possible interaction of the electromagnetic magnetic field on the EMG sensors.

Motion sensors were placed over L1 and S1 vertebrae of a subject while two EMG sensors were positioned on the erector spinae muscle, one 30 mm left and the other 30mm right from the L3 vertebra. Resting EMG was collected with the electromagnetic devices turned on and off for 10 seconds. Average and RMS values of the EMG raw data from resting signals were compared and no alterations in the EMG signal were found in the 3 tests performed as is shown in Table 3-4.

These suggested that there were no significant interactions between the devices.

Table 3-4: Results of the Motion tracking device interference in EMG data tests

SENSOR	Motion device	Test 1		Test 2		Test 3	
		ON	OFF	ON	OFF	ON	OFF
1	Average activity (mV)	-0.0088	-0.0088	-0.0252	-0.0252	-0.0173	-0.0172
	RMS (mV)	0.0115	0.0120	0.0258	0.0258	0.0234	0.0234
2	Average activity (mV)	0.0005	0.0006	0.0197	0.0198	0.0097	0.0097
	RMS (mV)	0.0163	0.0165	0.0343	0.0377	0.0208	0.0212

3.4.3 Force platform

3.4.3.1 Features

Kistler™ force platforms (Type 9281B, Kistler™ AG, Winterthur, Switzerland) is used for measuring forces, moments and coordinates of the force applied on the plate. The plate is formed by a metal plate with four piezoelectric transducers in each corner become charged when they are displaced in result of a load over the plate. The electrical charge is then converted to an analogue output that is connected to an amplifier. The amplifier is connected to the computer via a digital acquisition system that amplifies and converts the analogue signal into a digital signal. The force platform was operated using

dedicated software (Kistler BioWare software). The software permits control of different features such as the sensitivity, sampling frequency, length of the acquisition. Output data can be connected to any analogue acquiring system and processed by different data analysing software. The metal plate is provided with four threaded holes that can be used to fix objects on it and they are used to firmly attach the swinging chair on the force platform.

3.4.3.2 Validation tests

The manufacturer of the Kistler™ force platform recommended zeroing the force platform before each trial. In order to do that the swinging chair needed to be detached from the platform after each trial. This would have increased the time taken to complete the experiment because the person would have needed to get off the chair in order to remove and replaced between trials. The solution adopted to avoid this time commitment involved recording the static force due to the weight of the person and weight of the chair on the force platform just before starting the trials. The plate was zeroed before each trial and forces involved in the experiment recorded. Thereafter, the pre-recorded static forces were added to the measured force in the data analysis. A Validity test was performed to evaluate the effectiveness of this solution I. During the test the chair was tilted and released, and data from force platform acquired. The test was divided in two trials, in the first (trial A) the force platform was zeroed and then the swinging chair placed on it and the test performed; in the second (trial B) a static measure of the chair was taken and then the force platform was zeroed with the chair on it. Each trial was repeated 3 times and results averaged. Value of the X, Y and Z components of the forces were collected during all the tests. Difference between test A and test B were calculated for maximum and minimum value of the 3 components of the

force. The difference between test A and B was calculated for all variables. Differences, as showed in Table 3-5, were less than 3.5 N for all components of the forces.

Table 3-5: Results of the force platform validation tests

		Mean test A	Mean test B	Difference
Force X (N)	Min	-39.59	-39.26	0.33
	Max	-177.88	-174.66	3.22
Force Y (N)	Min	66.47	67.92	1.45
	Max	54.82	58.30	3.48
Force Z (N)	Min	522.57	522.81	0.24
	Max	517.12	517.39	0.26

3.4.4 Data acquisition board

3.4.4.1 Features

Data Translation module (DT9803-16SE-BNC) is a multi-functioning data acquisition system. There are 16 single ended or 8 differential analogue inputs which can be connected by BNC connectors or through a 37-pins D-sub connector. The device has programmable gains ($G=1, 2, 4, 8$), 16-bit resolution, 8 fast digital inputs, 8 digital outputs, 1 dynamic digital output and 2 counter/timer (16-bit). The system is linked to a computer by USB port. This module was selected because different devices could be connected at the same time. Moreover the devices connected could also be synchronised with other devices which were connected to the same computer through in other way (e.g. through the serial port). Here, each channel of the EMG device was connected to the BNC connector; the output of the Force platform was connected to the 37-pins D-sub connector. The module is controlled by dedicated software (Measure Foundry).

3.4.4.2 Synchronisation validity test

Measurements made from different equipment needed to be synchronised for data analysis. Data from force platform and EMG system were transferred to the computer through the data translation acquisition system, which allowed synchronization of the data. The data from the motion system were sent to the computer from the serial port. Synchronisation between data from Data Translation system and motion data was conducted through Measure Foundry software.

Tests were completed to determine the quality of the synchronisation between serial port data and data from the data acquisition system. In the test a motion sensors from the electromagnetic system was placed over the moving part of a plastic goniometer and an electrogoniometer sensor was placed over the plastic goniometer and connected to the computer through the data translation. The moving part of the goniometer was rotated and data were acquired from the motion sensor and the electrogoniometer. Cross correlation between motion sensor and electrogoniometer data was performed in order to evaluate time lag between the two series of data. Time lag with the highest correlation was reported. For all the trials the highest correlation was at time lag 0, indicating that the devices were successfully synchronised. Results of the test are shown in Table 3-6.

Table 3-6: Synchronization validity tests results

	Test 1	Test 2	Test3
Correlation coefficient	0.9987	0.9992	0.9991
Time lag (s)	0	0	0

4 STIFFNESS PROPERTIES OF THE TRUNK IN PEOPLE WITH LOW BACK PAIN

4.1 Introduction

The stiffness of the trunk has been suggested as an important factor that contributes to postural control of the back in different activities (Cholewicki et al., 1997, Gardner-Morse and Stokes, 1998, Gardner-Morse et al., 1995). Trunk stability is achieved with different structures assuming different roles (Panjabi, 1992). The passive stiffness of the bones and ligaments of the spine are not sufficient to control the posture of the trunk (Cholewicki and McGill, 1996, Crisco and Panjabi, 1992, Crisco et al., 1992). Stability is largely achieved by the stiffness of the muscles and modulated by the neural system (Crisco and Panjabi, 1992, Gardner-Morse and Stokes, 2001).

LBP and spine injuries can affect the viscoelastic properties of spinal structures and influence the trunk stability (Panjabi, 1992). Intrinsic alterations in viscoelastic properties of the spinal structures have been evaluated in cadaveric studies, which showed that motion segments with degenerated discs exhibited decreased stiffness and damping coefficients (Keller et al., 1987). Consequently, these segments were found less stable, with a higher creep rate compared to the less degenerated segments. Alterations in viscoelastic properties of the spine were also observed in-vivo. For instance, the postero-anterior lumbar spine stiffness was examined in LBP and healthy participants by Latimer et al. (1996b). They employed a device that delivered a force to the L3 level spinous process while the subject was lying on a table in prone position. Stiffness was calculated as function of force and displacement, and it was shown that LBP was found to have increased stiffness in the trunk. Symptomatic participants were also re-tested after that the pain was resolved by more than 80%, and a significant

decrease in the stiffness was shown in this second test (Latimer et al., 1996b), highlighting the association between increased stiffness and pain. An improved version of this device was used by Colloca and Keller (2001) to evaluate the posteroanterior stiffness at different levels of the trunk and similar results were found, showing that LBP subjects with high recurrence of pain had increased stiffness in the spinous processes in comparison to healthy subjects or subjects with less frequent LBP. Shum et al. (2013) used same method to evaluate the effect of spinal mobilization and they found decreased stiffness and pain in LBP subject after spinal mobilization therapy.

Hodges et al. (2009) investigated viscoelastic parameters in a semi-upright sitting position, and they found an increase in the spine stiffness for LBP subjects and a decrease in damping coefficient. The researchers suggested that the alterations in the spine stiffness may be due to increased trunk muscle activity to protect spinal structures. This may be a mechanism to compensate for the reduced damping as a result of the physiological changes in passive structures.

Balance control of the trunk in response to a perturbation has been investigated using centre of pressure trajectory. The findings showed that there was an increase in the postural sway in LBP subjects (van Dieen et al., 2010, Cholewicki et al., 2000b, Preuss et al., 2005). However, they did not determine how the balance mechanism was related to the viscoelastic properties of the trunk.

A limitation of previous studies is the lack of ecological validity as the participants were prone or semi-upright sitting, where the effect of gravity is altered and where there was no dynamic response to a perturbation (Brown and McGill, 2009, Gardner-Morse and Stokes, 2001, Hodges et al., 2009, Latimer et al., 1996c, McGill et al., 1994). Standing position has been also used to evaluate mechanical properties in healthy subjects, and due to the contributions of the lower limbs, the effectiveness of the trunk in maintaining balance cannot be ascertained (Moorhouse and Granata, 2005).

In order to remove the influence of the lower limbs, the viscoelastic properties of the trunk would be examined in a sitting position in this experimental study, while the participants will try to regain a balanced position after been tilted on a swinging chair. This would simulate common activities such as sitting on a bus or in a car. Motion and moment data will be used to determine the viscoelastic properties of the trunk in a dynamic situation.

The aim of this study was to employ a second order linear model to derive the viscoelastic parameters of the trunk while a subject was performing a balancing task, and to examine the differences in these properties between healthy and LBP subjects. It was expected that participants would alternatively flex and extend the trunk to find the balanced position; the relation between trunk displacement and external moment acting on the trunk would allow us to derive the viscoelastic properties through the model. It is hypothesized that LBP subjects would have larger trunk stiffness compared to healthy subjects.

4.2 Methods

4.2.1 Participants

Thirty healthy participants without a history of LBP, by self-report and twenty-four participants with sub-acute (>6 weeks) LBP (Savigny et al., 2009) were recruited for the study. Exclusion criteria for all participants included presence of ankylosing spondylitis, fractures/dislocation of the spine or hips, history of spinal or hip surgery, pregnancy, neurological disorders, cancer and osteoporosis. The rate of severity of pain of the LBP participants was recorded using a visual analogue scale (VAS) and the functional ability evaluated by Oswestry Disability Questionnaire (Fairbank and Pynsent, 2000) (Appendix A1).

The characteristics of the participants of the two groups are summarised in Table 4-1. No significant differences were found between the two groups for age, weight, height and BMI.

Table 4-1: Participants' characteristics (no significant differences between groups)

	Healthy participants (n=30) mean \pm SD	LBP participants (n=24) mean \pm SD
Gender	10 males 21 females	12 males 11 females
Age (yr)	31.73 \pm 8.10	36.83 \pm 11.56
Height (m)	1.676 \pm 0.980	1.689 \pm 0.840
Weight (kg)	63.89 \pm 13.33	68.96 \pm 11.64
BMI (kg/m ²)	22.53 \pm 2.67	24.09 \pm 3.09
VAS score (scale 0-10)	N/A ^A	3.80 \pm 1.02
Oswestry score (scale 0-100)	N/A ^A	19.83 \pm 8.94

^Anot applicable

4.2.2 Equipments

A custom-made chair was built which was restricted to swing in the sagittal plane; it had foot and leg support in order to restrict the knee and ankle to a 90° angle. The base of the chair was mounted onto a force platform (Type 9281B, KistlerTM). The chair was built from wood as metal would interfere with the electromagnetic field generated by the transmitter of the motion tracking system. The swinging mechanism was provided by two low friction ball bearings. Mechanical stops prevented the chair from swinging more than 20° backward and forward. The chair was provided with adjustable belts and footrest in order to restrict movements of the lower body and to reduce the risk of falling from the chair. In this manner the angular displacement of the lower limbs of the participants was equal to the displacement of the chair. The chair was also designed with a sliding mechanism to allow the participant to balance in the resting position.

The movement of the lumbar spine was measured using a three-dimensional motion track system (3SPACE FASTRAK®, Polhemus Inc.) recording at 40 Hz. Previous work showed that the error due to relative motion between the sensors placed on the skin and the vertebrae was acceptable, and less than 8% of the spine motion observed (Yang et al., 2008). Two sensors were placed on the participants' back, one at the sacrum level and one at the first lumbar vertebral level. One further sensor was placed on the chair to track its rotation, which was also the rotation of the lower limbs. These data were used to derive the trunk angular displacement and for the trunk moment calculation. The chair was placed over the force platform recording at 150 Hz, in order to determine the loads that acted on the system formed by the chair and the participant. Measure Foundry (Data Translation Inc.) software was used to synchronized data acquisition of the devices and to integrate motion and force data. MATLAB® (R2007b, MathWorks Inc.) was used for data resampling and analysis and SPSS (SPSS: An IBM Company) was used for statistical analysis.

4.2.3 Protocol

A schematic diagram of the participant sitting on the swinging chair is shown in Figure 4-1. Participants were strapped to the chair to immobilise the lower limb and pelvis, with their arms folded across the chest facing forward. The height of the feet support was adjusted to allow the participant to sit in a comfortable position. Body landmarks (pelvis and spine landmarks) were digitized using a wooden probe attached to one motion sensors and data were used to define hip joint center as the caudal endpoint of the trunk according with Zatsiorsky (2002). A standard familiarisation protocol allowed each participant to become familiar with the testing protocol and motion. Firstly the participant was tilted twice backwards and forwards by the researcher in a controlled manner to

show the range of motion (ROM) of the chair; then the balanced position (with the chair parallel to the floor) was shown to the subject.

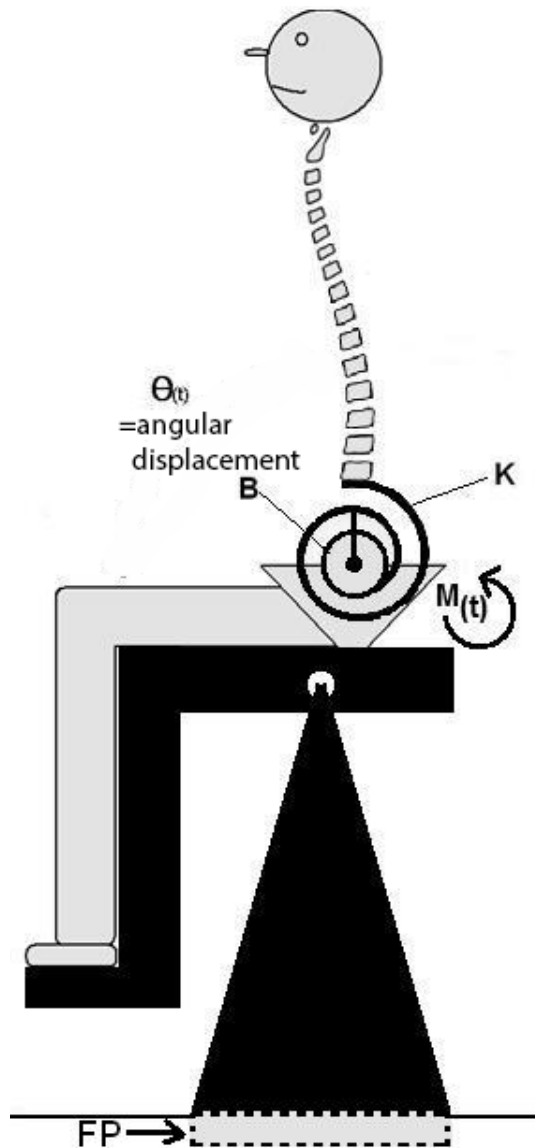


Figure 4-1: Schematic drawing of the experiment, showing the swinging chair and the participant. Spring (K) and dashpot (B) represent the elastic and viscoelastic properties of the trunk. $M(t)$ and $\theta(t)$ are the moment acting at the trunk and the corresponding angular displacement. FP indicates the location of the force platform.

Thereafter, the researcher tilted the chair into an angle between $0-20^\circ$ and following release, asked the participant to return to the balanced position and maintain it for 5s. This familiarization protocol ended when the participant was able to find the balanced

position and to hold it for at least 5 seconds for three repetitions. All participants were able to complete the familiarization protocol. After the familiarization protocol, the force plate was zeroed and the chair tilted 10° forward using the handle. The chair was released without warning the participant and they were asked to achieve a steady balanced position. Each participant had three attempts to reach the balanced position. The trial was considered successful when the participant was able to reach the balanced steady position without reaching the mechanical stops and maintaining it for 1 second with a maximal oscillation of $\pm 1^\circ$. The testing protocol was then repeated with the chair tilted 20° forward.

4.2.4 Analysis

Moment and angle data were processed with a fifth order Butterworth low pass filter using a 2 Hz cut-off frequency. Inverse dynamic equations were used to derive the moment acting at the hip joint using data from force platform and motion tracking system (Figure 4-1). The hip joint center was located using the digitized landmarks points. Mass, centre of mass and inertial properties were calculated using the nonlinear regression equations from Zatsiorsky (2002).

The trunk moment M_T (Figure 4-2) was derived the following equation:

Equation 4-1

$$M_T = I_H \varepsilon_S - F_z d_1 + F_y d_2 + m_L g d_3$$

where:

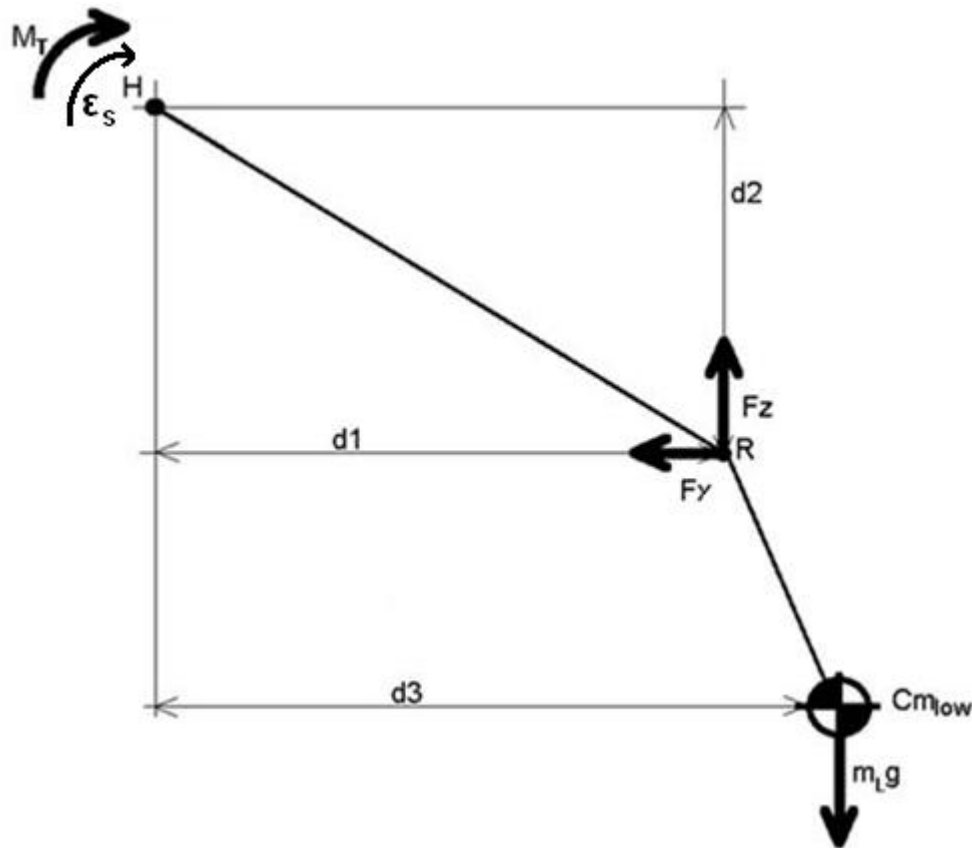
R =centre of rotation of the swinging chair; H =Hip joint center;

Cm_{LOW} =centre of mass of lower limbs and swinging part of the chair;

d_1 =distance between H and R in y axis; d_2 =distance between H and R in z axis;

d_3 =distance between H and Cm_{LOW} in y axis; ε_S =angular acceleration of the hip;

I_H = total inertia (sum of the inertia of left and right foot, shank, thigh and swinging part of the chair);



Trunk dynamic properties about the hip were derived using a second order linear model (Figure 4-1)(D'Azzo and Houpis, 1988). The following equation describes the model:

$$\ddot{\theta}(t)J + \dot{\theta}(t)B + \theta(t)K = M_T(t)$$

where: $\ddot{\theta}(t)$ =angular acceleration; J =moment of inertia; $\dot{\theta}(t)$ =angular velocity;

$\theta(t)$ =angular displacement; K =trunk stiffness coefficient about the hip; B =trunk damping coefficient about the hip; $M_T(t)$ =trunk moment.

Trunk angular displacement was calculated as the relative motion between the sensor on the chair and the sensor in L1. Data showed that the trunk motion was mainly in the sagittal plane, the ROM in the frontal and transverse plane was less than $\pm 1^\circ$ for all participants. Thus the modelling was confined to the sagittal plane only. The trunk displacement was numerically differentiated over time to calculate trunk angular velocity and acceleration. The standard least square method was used to estimate the viscoelastic parameters. For both trunk angular displacement and moment the balanced position was defined as zero. Maximal peak to peak values (from maximum positive peak to maximum negative peak) during the trial were reported for trunk angular displacement and for the trunk moment. Trunk flexion was considered when angular displacement was positive while trunk extension was with negative angle, similarly, a positive moment was considered a flexor moment, while a negative moment was extensor moment. Onset of chair movement (T1) was defined as the instant when the chair angle deviated more than 3 SD from the mean of the chair angle calculated for 0.2 s time window prior to the release. Return of the chair to the balanced position was defined as the instant when the chair angle deviated less than 3SD from the mean chair angle calculated for 0.2 s time window in steady state.

4.2.5 Statistics

Independent T-test was used to compare the difference between LBP and healthy participants in the mean stiffness, damping and moment of inertia coefficients, the peak to peak trunk angular displacement and the hip moment. Statistical analysis was also performed to evaluate the effects of initial tilting angle and gender on the results. The

significance level was set at 0.05. Coefficient of determination (R-square) was used to evaluate the goodness of the fit of the second order linear model.

4.2.6 Reliability

Five LBP and five healthy subjects were re-tested within a week from the first test to evaluate the reliability of the data provided by the experimental model (Appendix A2). Pearson product moment coefficient was used to determine the reliability for the stiffness, damping and moment of inertia coefficients.

4.3 Results

As soon as the chair was released, the trunk flexed to decelerate the chair rotation (Figure 4-3).

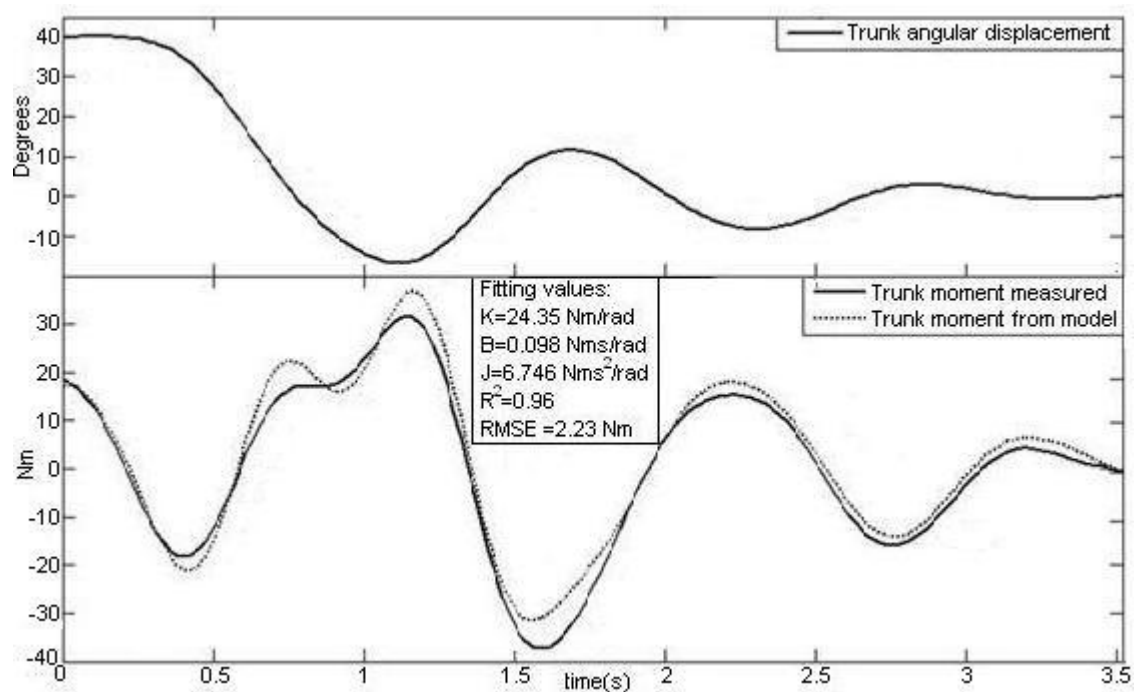


Figure 4-3: The angular displacement and moment of the trunk during a 20° tilt of an experimental subject. The moment predicted by the second order model and inverse dynamics are compared showing the goodness of fit. The moment parameters - moment of inertia (J), stiffness (K), damping (B), determination coefficients (R^2), RMSE- are shown in the figure.

This was followed by alternate flexion and extension of small magnitude until the balanced position was re-achieved.

All participants followed a similar pattern. There was no significant difference in the mean of the ROM of the trunk, calculated as the peak to peak value of the trunk displacement, between the healthy and LBP groups. The mean range increased significantly when the tilt angle of the chair was increased from 10° to 20° (25.44° to 33.60° for healthy group and 25.50° to 34.07° for LBP group, Table 4-2). The trunk moment also oscillated until the balanced position as shown in Figure 4-3; The average for the peak to peak of the trunk moment tended to be higher for the LBP group for both trials, but the difference was not significant (Table 4-2).

Table 4-2: Mean and Standard deviation of the stiffness, damping and moment of inertia coefficients, trunk ROM, and R-square coefficient for both the group and both the trials (10° and 20° initial tilt angles)

	10° TILTING		20° TILTING	
	HEALTHY	LBP	HEALTHY	LBP
	mean ± SD	mean ± SD	mean ± SD	mean ± SD
Trunk ROM (°)	25.44 ± 10.10	25.50 ± 8.73	33.60 ± 10.33	34.07 ± 9.85
Trunk moment (Nm)	27.41 ± 7.46	35.22 ± 20.54	39.18 ± 14.04	41.90 ± 10.53
Moment of inertia (Nms/rad ²)	5.72 ± 1.47	6.23 ± 1.46	5.76 ± 1.33	6.39 ± 1.24
Stiffness (Nm/rad)	35.28 ± 10.22*	49.33 ± 6.34*	35.17 ± 7.69*	47.52 ± 8.09*
Damping (Nms/rad)	0.57 ± 0.78	0.70 ± 0.85	0.34 ± 0.31	0.66 ± 0.67
R-square coefficient	0.85 ± 0.08	0.87 ± 0.08	0.88 ± 0.09	0.89 ± 0.07

*Indicates significant differences between LBP and healthy groups at a given tilt angle (p-value < 0.05)

There was a high correlation ($r > 0.92$ for all subjects) and a low RMSE (1.9 Nm for both groups) between the moment determined by the inverse dynamics and that predicted by the second order model using the viscoelastic properties derived, demonstrating the

goodness of the model proposed. In the test-retest analysis, the stiffness and moment of inertia data were found to be highly reliable ($ICC > 0.9$) Reliability was low ($ICC < 0.4$) for the damping coefficient, but this was due to the value of this coefficient being almost zero for both subject groups. The model revealed that stiffness coefficient was significantly increased for the LBP group ($p=0.000$ for both initial angles 10° and 20° , Table 4-2) However, no significant differences in the variables were found between male and female participants for both healthy and LBP groups.

4.4 Discussion

The aim of the experiment was to determine the mechanical behavior of the trunk while a subject was sitting on an unstable chair and attempting to maintain a balanced position after being tilted. The strategy used to return the trunk to a balanced position quantified the different mechanisms utilized by participants with and without LBP. The external trunk moment determined by inverse dynamics represented the net moment produced by the muscles to achieve moment equilibrium. Previous research has indicated that the viscoelastic properties of the trunk are largely modulated by muscle activation and to a smaller extent the passive moment generated by the joints (Brown and McGill, 2009, Cholewicki et al., 2000a, Colloca and Keller, 2001). Previous in-vitro experiments evaluated the viscoelastic properties without considering muscular activation (Keller et al. 1987). The present study analysed the trunk as a whole system in a weight-bearing position; this allowed us to determine the viscoelastic properties of the trunk in a realistic condition, including all the passive and active components (Nichols and Houk, 1976). This approach challenges the trunk function without the influence of the lower limbs and enables us to quantify possible differences between healthy and LBP participants in regard to their viscoelastic properties; furthermore, the current study considered the angular motion of the trunk and included the inertial effects

of trunk to provide the best estimation of trunk viscoelastic properties. These properties were derived by fitting the moment-data data with a second order linear model. The moment predicted by this model using the derived properties were compared with that determined by the inverse dynamic model. The goodness of fit was excellent ($r=0.92$), suggesting that the model accurately described the behaviour of the trunk.

Our findings showed that the response of the trunk was mainly elastic, as the damping coefficient was almost zero for all participants. The response to the perturbation was independent of the trunk velocity, a change of the external moment was followed by a proportional change of displacement; this means that trunk responded with a sudden movement to balance the external forces. This finding was in agreement with that of Brown and McGill (2009).

Passive stiffness of the trunk in flexion and extension, as reported by McGill et al. (1994), was 7.5-18.3 Nm/rad. In this study the trunk stiffness values were higher, about 35 Nm/rad for healthy and 49 Nm/rad for LBP subjects. This is because the current stiffness is not passive and includes active trunk muscular contractions required to maintain balance on a chair (Brown and McGill, 2009).

Trunk stiffness coefficient was found to be significantly increased in LBP participants; this was in accordance with previous research from Hodges et al. (2009) The increase in stiffness found in LBP participants may be explained as a compensatory strategy in order to reduce lumbosacral motion, instability and to minimise pain or the risk of further damage of the spine tissues during perturbation (Shum et al., 2005a, Shum et al., 2005b). In this study only the total trunk ROM is analysed and no differences are found between healthy and LBP participants, but differences may be found in the ROM of hip and lumbosacral joint and in their contribution to the total trunk ROM. Further studies are needed to evaluate the kinematics and the contribution of these joints to the trunk motion. As all participants were able to perform the task within the three attempts, it can

be stated that alterations in viscoelastic properties did not compromise the ability of LBP subjects in balancing on an unstable chair, but alterations were an effective compensatory strategy. The results of the experiment were not affected by the initial tilt angle: with 20° initial tilting the only difference between LBP and healthy participants was in the magnitude of the stiffness coefficient. Previous studies found increased muscular activity and co-contraction of agonist and antagonist muscles involved in the postural control of the trunk as results of LBP (Radebold et al., 2000, van Dieen et al., 2003). This may explain how the increasing in stiffness is achieved. Investigating the trunk muscular activity during unstable sitting should be carried out in the future in order to understand the mechanism behind viscoelastic properties alteration showed to be present in LBP subjects.

This experiment does not address the issue of whether the change in elastic properties of the trunk is due to LBP as a compensatory mechanism to minimise discomfort, or the change in properties is the cause of LBP. This needs to be clarified as this will affect how LBP should be managed. If the stiffness is due to pain, relief of the symptoms should be provided, and the clinicians should examine if there is a corresponding decrease in stiffness. If stiffness is the cause of the pain, exercise therapy may have a role in managing the condition. In this case, it is important that the patients should keep active and not develop fear of motion, which may lead to further increase in stiffness and pain.

Another limitation of the study was that viscoelastic properties were evaluated only in the sagittal plane because the motion of the swinging chair was restricted to this plane of motion. Future study may use a modified version of the chair which can move in all directions, allowing estimation of the trunk's viscoelastic properties in three dimensions. This may provide more information about compensatory strategies of the trunk in complex loading situations. Further research will also be required to clarify muscular

activation mechanism and to evaluate possible alteration in joint loading due to kinematics changes that may lead to worsening of the symptoms.

4.5 Conclusions

The biomechanical response of the trunk to a swinging chair can be adequately described by a second order linear model. It was shown that the response was mainly elastic for all participants, and LBP participants exhibited an increase in the stiffness of the trunk. Such increase may be a compensatory strategy to reduce pain and risk of further injuries, although it may also be cause of LBP. Further research is needed in order to clarify the cause and effect relationship, and the role of the muscular activity in the compensatory strategies.

5 TRUNK DYNAMIC STABILITY DURING UNSTABLE SITTING IN PEOPLE WITH LOW BACK PAIN

5.1 Introduction

Low back pain (LBP) is one of the most widespread pathological conditions and often related to impairment or alteration of the trunk mobility in common activities (Savigny et al., 2009, Strand and Wie, 1999). An understanding of the relationship between LBP, trunk kinematics and its dynamic stability in a certain task will help us evaluate the possible strategies that LBP subjects may adopt, and the effectiveness of these strategies.

Trunk kinematics can be evaluated in standing and sitting tasks, but the sitting position has the advantage of excluding the contribution of the lower limbs, focusing exclusively on the trunk behaviour. In particular, trunk motion in sitting has been evaluated for everyday tasks such as picking up an object (Shum et al., 2007a), and putting on a sock (Shum et al., 2005b). It was shown that hip and lumbar spine motions were decreased in the sagittal plane with a corresponding increase in motion in the other planes (Shum et al., 2005b, Shum et al., 2007a). LBP subjects were also found to need more time to complete the task indicating a compromised performance of the task. Postural control during unstable sitting (Van Daele et al., 2009, Radebold et al., 2001), and the response to a perturbation (Cholewicki et al., 2000a, Preuss et al., 2005, Radebold et al., 2001, Van Daele et al., 2009), where subjects were asked to maintain a steady position as much as they could. Dynamic stability was examined using the centre of pressure sway, and it was found that LBP subjects exhibited an increase in the sway that might imply a poorer balancing performance due to pain (Cholewicki et

al., 2000a, Preuss et al., 2005, Radebold et al., 2001). However, in studies where postural stability were evaluated in standing (Maribo et al., 2012) and stable sitting (van Dieen et al., 2010), different results were observed. The centre of pressure motion was not significantly increased in LBP subjects, and no correlation were found between CoP, pain intensity, fear of pain and physical functions (Maribo et al., 2012).

This study used a different approach compared to the earlier work (Cholewicki et al., 2000a, Preuss et al., 2005, van Dieen et al., 2010). The centre of pressure sway was a useful indicator of dynamic stability but it did not describe the trunk motions and how stability was achieved. In our study, the kinematic response to a perturbation would be modelled with a second order equation (Bajd and Vodovnik, 1984, Ditroilo et al., 2011a). This would allow us to characterise the damping and natural frequency characteristics of the response, and to evaluate the subjects' performance in terms of their ability to regain balance (using the duration and accuracy of regaining balance). We would also study the kinematics of the spine and the hip so as to understand their relative contributions to balance control. We hope this experimental approach could provide new insights into the biomechanics of the spine in unstable sitting and to re-examine some of the discrepancies in the findings of previous work.

Previous work showed that alterations in lumbar lordosis were observed in radiographic images of LBP subjects (Jackson and McManus, 1994, Tsuji et al., 2001). Scannel and McGill (2003) showed the changes in the trunk tissue strain as a result of the alteration of the lordosis curvature. It is therefore possible that the pain-related changes in posture would have an effect on dynamic balance control. This hypothesis needs to be clarified, and would be examined in this study. The purpose of this study was to evaluate the dynamic stability and kinematics of the lumbar spine during unstable sitting, and to model the response using a second order equation and determine the differences in the model parameters between healthy and LBP subjects.

5.2 Methods

5.2.1 Participants

(This paragraph is similar to the paragraph 4.2.1, it has been left in this section to preserve the integrity of the article as it was sent to publication)

Thirty healthy subjects without history of LBP and twenty-four subjects with sub-acute (6-12 weeks) LBP (Savigny et al., 2009) were recruited for the study. Exclusion criteria for all subjects included presence of ankylosing spondylitis, fractures/dislocation of the spine or hips, history of spinal or hip surgery, pregnancy, neurological disorders, cancer and osteoporosis. The severity of pain of the LBP subjects was recorded using a visual analogue scale (VAS) and the functional ability evaluated by Oswestry Disability Questionnaire (Fairbank and Pynsent, 2000) (Appendix A1). Subjects' characteristics are summarised in Table 5-1 and are not significantly different. Informed consent was obtained from all subjects (Appendix A1), and the study was approved by the Ethics Committee of the University of Roehampton.

Table 5-1: Participants' characteristics (no significant differences between groups)

	Healthy participants (n=30) mean \pm SD	LBP participants (n=24) mean \pm SD
Gender	10 males 20 females	12 males 12 females
Age (yr)	31.73 \pm 8.10	36.83 \pm 11.56
Height (m)	1.676 \pm 0.980	1.689 \pm 0.840
Weight (kg)	63.89 \pm 13.33	68.96 \pm 11.64
BMI (kg/m ²)	22.53 \pm 2.67	24.09 \pm 3.09
VAS score (scale 0-10)	N/A ^A	3.80 \pm 1.02
Oswestry score (scale 0-100)	N/A ^A	19.83 \pm 8.94

^Anot applicable

5.2.2 Equipment

A custom-made chair was built, which was restricted to swing in the sagittal plane; the chair was provided with foot and leg supports in order to restrict the knee and ankle to a 90° angle. The chair was built from wood as metal would interfere with the electromagnetic field generated by the transmitter of the motion tracking system. Mechanical stops prevented the chair from swinging more than 20° backward and forward. The chair was also provided with adjustable belts and footrests in order to restrict movements of the lower body and to minimise the risk of falling from the chair.

The movement of the lumbar spine was measured using a three-dimensional motion track system (3SPACE FASTRAK®, Polhemus Inc., Colchester, VT) recording at 40 Hz. Previous work showed that the error due to relative motion between the sensors placed on the skin and the vertebrae was less than 8% of the gross spine motion (Yang et al., 2008), and this was considered to be acceptable. Two sensors were placed on the subjects' back, one at the sacrum level and one at the first lumbar vertebral level. One further sensor was placed on the chair to track its rotation, which also defined the rotation of the lower limbs.

5.2.3 Protocol

Subjects were strapped to the chair with the lower limbs and pelvis immobilised, and with their arms folded across the chest facing forward. The height of the feet support was adjusted to allow the participant to sit in a comfortable position (Figure 5-1). The subjects were allowed to swing freely on the chair to get familiarised to the testing condition. Two tilt angles were tested in random order, 10° and 20°. The researcher tilted the chair into the required tilt angle, and the chair was released without warning to the subjects.

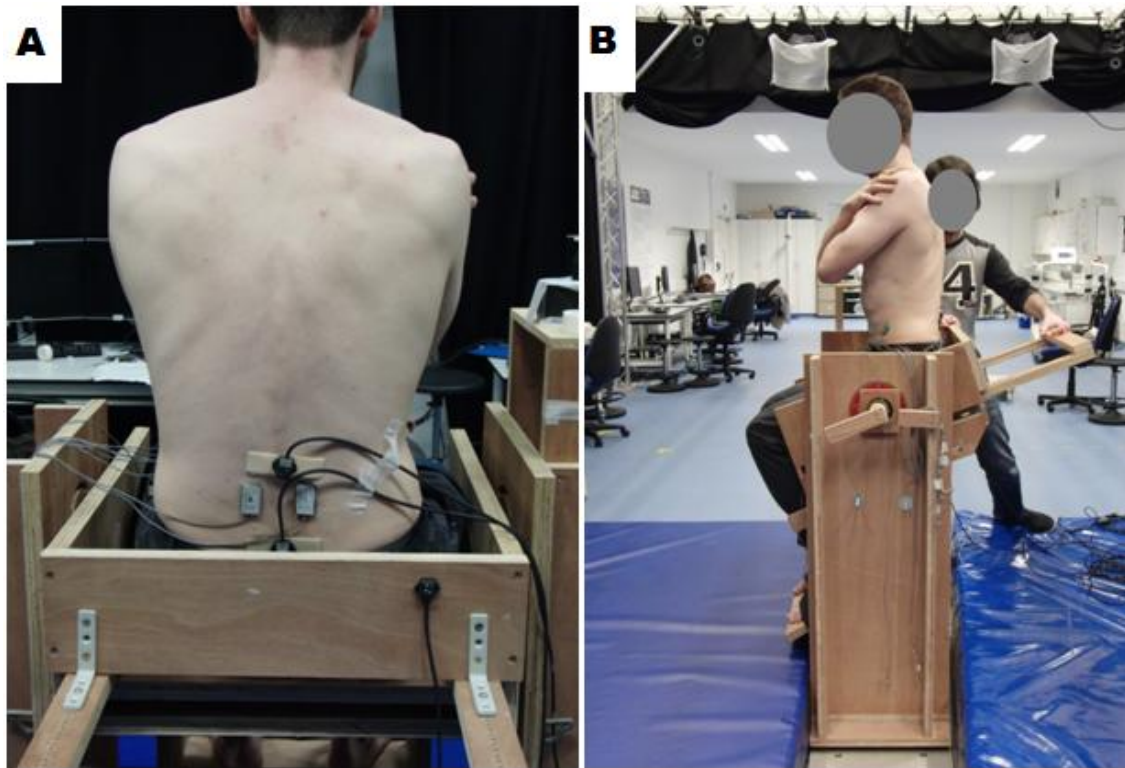


Figure 5-1: The experimental set-up of the participant on the swinging chair; A is Back view and B is Side view

They were then asked to achieve a steady balanced position as soon as possible. The trial was terminated when the participant was able to regain stable balance. This was considered to be the case when the oscillation of the chair could be maintained for less than $\pm 1^\circ$ for more than one second. The testing protocol was repeated three times for each tilt angle.

5.2.4 Posture evaluation

The curvature of the spine was measured in upright standing using the method proposed by Singh and co-workers (Singh et al., 2010). A probe attached to an electromagnetic tracking sensor (3SPACE FASTRAK®, Polhemus Inc., Colchester, VT) was used to digitally track the spine curvatures. The magnitude of the thoracic kyphosis

and lumbar lordosis angles were determined. The reliability of this measurement method was excellent. The mean intraclass correlation coefficient (ICC) for kyphosis measurement was 0.93 with a mean absolute error of 1.57°, whereas for lumbar lordosis measurement, the mean ICC was 0.98 and an SEM of 1.51° (Appendix A3).

5.2.5 Kinematic Analysis

The Cardan angle (Nigg and Herzog, 1994) was used to evaluate the relative motion between the two segments. Data from the electromagnetic sensors were used to obtain a rotation matrix for each sensor and then the Cardan angles were derived (the sequence of rotations was flexion/extension, followed by lateral bending or abduction/adduction and finally axial rotation). The hip angle is derived from the orientation between the pelvis and lower limbs, and the lumbar angle that between L1 and the pelvis. In this research, the hips on the left and right sides are assumed to move together because the participant's thighs were fastened onto the chair.

Hip and spine angular ROM were calculated as the differences between the maximum positive peak and the maximum negative peak; spine/hip ratio was calculated as the ratio between the spine and hip angular displacement values at the 1st (PK1) and the 2nd (PK2) peaks of the chair angle (Figure 5-2). As shown in Figure 5-2, there were negligible motions in the coronal and horizontal planes. Therefore, all analyses were performed in the sagittal plane only.

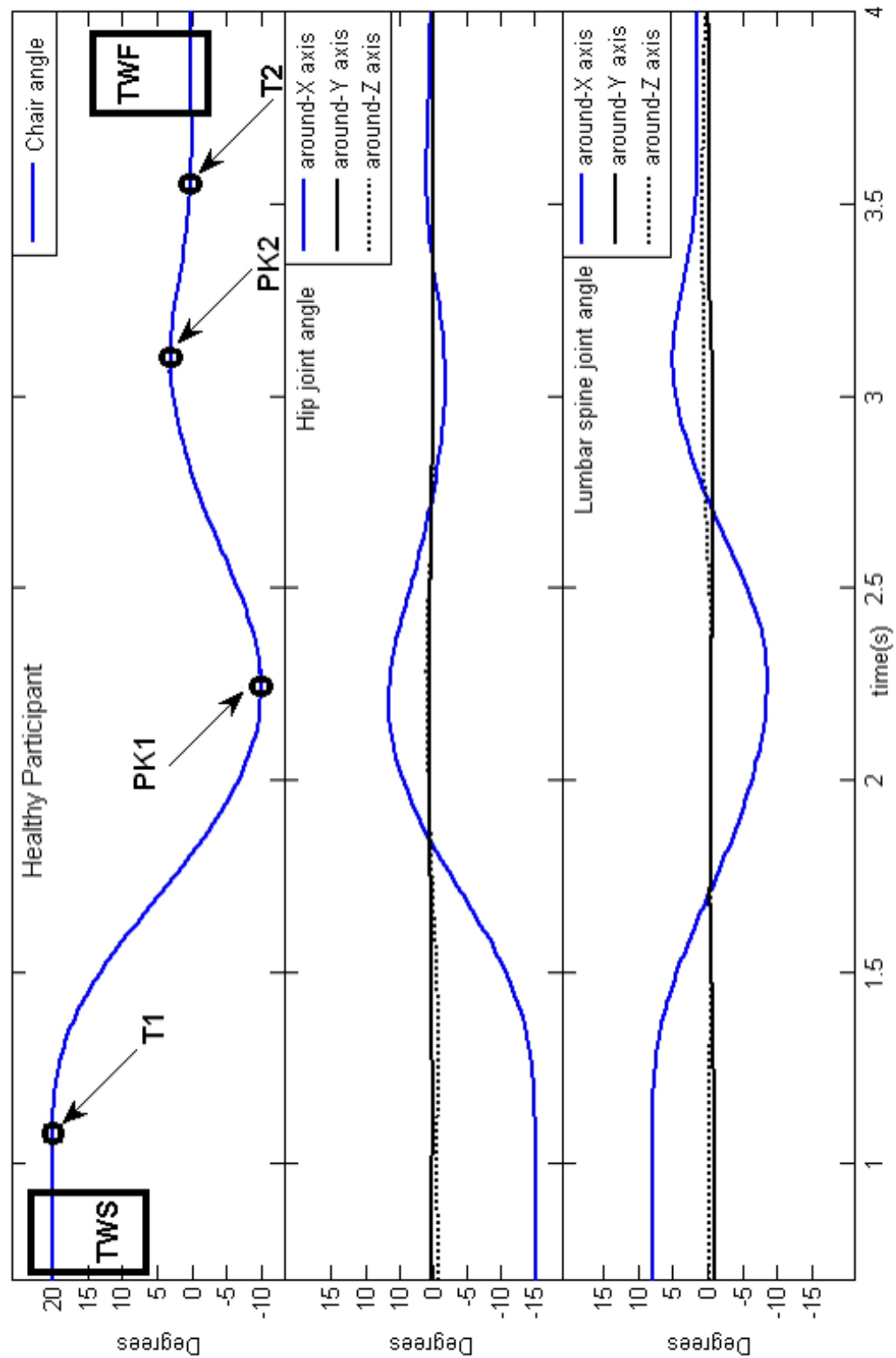


Figure 5-2: An example of the kinematics of the lumbar spine and hip in a healthy subject.

5.2.6 Dynamic stability analysis

The onset of chair movement (T1) was defined as the point where the chair angle deviated by more than 3 standard deviations (SD) from the mean of the chair angle (TWS) prior to release (Figure 5-2). Return of the chair to stationary point (T2) was defined as the point where the chair angle deviated less than 3SD from the mean chair angle (TWF) in a steady state (Figure 5-2). The balancing error was calculated as the difference between the ideal value of the chair angle for the steady balanced position (0°) and the steady balanced position achieved by the participant; for example, Figure 5-2 shows that the participant reached the steady position with a seat angle of -1.5° , and so the balancing error was 1.5° .

The angular displacement of the chair represents how the participant controlled the swinging to regain balance. In this study the participant was considered as a system that controlled the angular displacement of the chair. In order to evaluate the system response, the chair displacement was fitted with a second order system in response to a step input (the chair tilt) (D'Azzo and Houpis, 1988).

Equation 5-1

$$Y(t) = Y(0) * \frac{1}{\sqrt{1 - \zeta^2}} * e^{-t\zeta\omega_n} * \left\{ \sin \left[\left(\omega_n * \sqrt{1 - \zeta^2} * t \right) + \text{atan} \left(\frac{\sqrt{1 - \zeta^2}}{\zeta} \right) \right] \right\}$$

Where:

$Y(t)$ = chair angle curve; t = time; $Y(0)$ = chair angle at time 0;

ω_n = natural frequency (rad/s); ζ = damping ratio.

This type of analysis permitted evaluation of how the overall system deal with the perturbation and it would allow us to examine if LBP might alter the system response.

The standard least square method was used to estimate the model parameters, i.e. the natural frequency and the damping ratio. R-square (coefficient of determination) was used to check the goodness of the modelling. Figure 5-3 is an example of the fitting is shown from one healthy and one LBP participant performing the experiment with 10° initial trial.

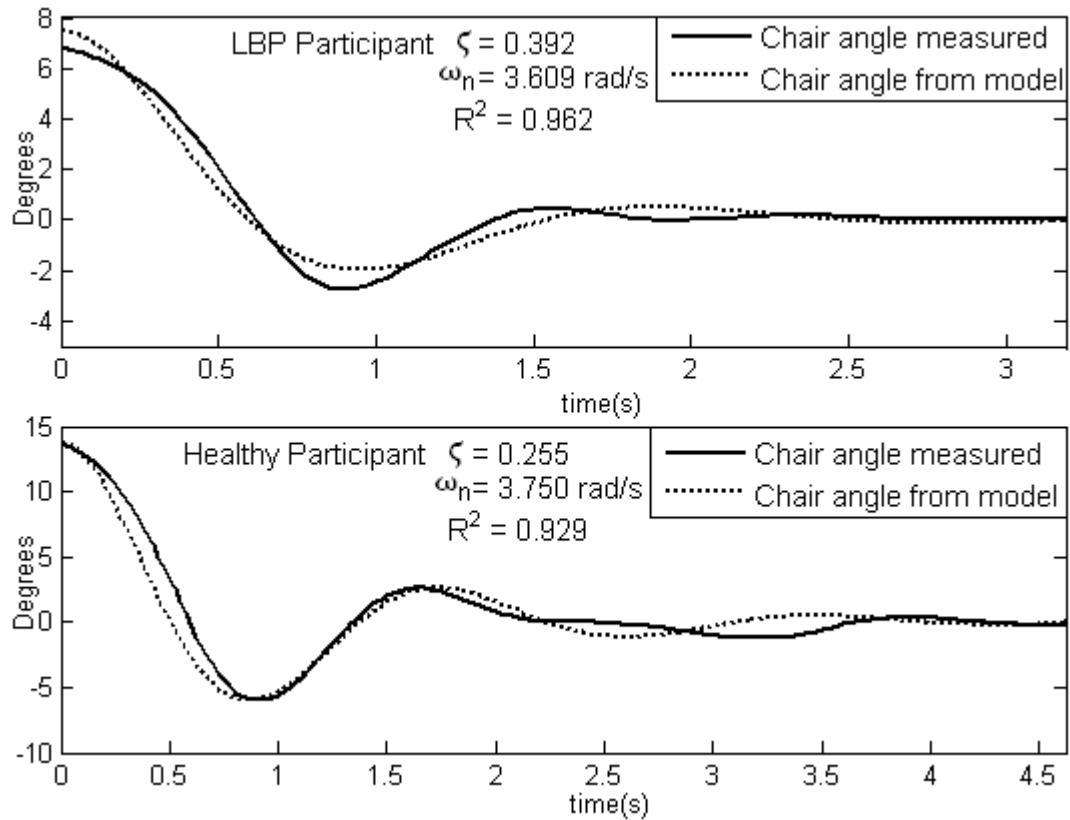


Figure 5-3: Comparison of the experimental data (chair angle vs time, and model parameters) between LBP and healthy participants.

5.2.7 Statistics

Multivariate analysis of variance (MANOVA) was performed to detect possible significant differences between LBP and healthy subjects groups in the trunk kinematics data (hip and spine ROM angles, spine/hip ratios, balancing error, time to regain balance, spine curvatures, and the model parameters (the natural frequency and

damping ratio) of the equation. The Shapiro-Wilks tests and the Box's M tests were used to confirm that the assumptions of normality and homogeneity of variance of the statistical model were met.

Pearson correlation coefficients were used to evaluate correlation among the variables described above. For all statistical analyses, significant level was set at 0.05, and all analyses were performed using SPSS (SPSS: An IBM Company, USA) software.

5.3 Results

MANOVA showed that there were no significant differences in any variables between male and female subjects ($P>0.05$). The analysis was thus performed with the data from male and female subjects altogether.

As shown in the example of Figure 5-2, hip and spine flexed and extended to control the movement of the chair and to regain the balanced position. The movements for both joints were mainly in the sagittal plane. The ROM of both joints in the frontal and transverse planes of motion was smaller than $\pm 1^\circ$. Initial tilt angle (10° and 20°) affected the magnitudes of the chair and hip angles ROMs. This was expected as a larger tilt would lead to larger chair and hip angles.

The hip ROM was found to be higher than spine ROM and the spine/hip ratio was smaller than 1 for all subjects, indicating the greater role of the hip joint. Hip and spine joint moved in a similar manner for LBP subjects, flexing and extending to perform the exercise. Differences were found in the ROM, in particular hip ROM increased while spine ROM decreased in LBP subjects for both tilt angles ($p<0.05$, Table 5-2).

Means of 1st and 2nd peak spine/hip ratios were significantly decreased in the LBP group for both trials, apart from the ratio calculated at the 2nd peak for the 20° trial whereby the differences were not significant (Table 5-2).

The chair angle, as shown in the two examples in Figure 5-3, exhibited a similar pattern of a damped oscillator: it decreases its oscillation until it reaches a stable position. Results showed that all subjects controlled the chair this manner.

Table 5-2: Mean and Standard deviation for trunk kinematic variables (ROM values for spine and hip joints), dynamic stability variables (ROM values for chair angle, balancing error and length of the trial), model parameters (damping ratio and natural frequency) an and spine/hip ratio (1st and 2nd peak) for both the group and both the trials (10° and 20° initial tilt angles)

	10° TILTING			20° TILTING		
Initial tilt angle	Healthy	LBP		Healthy	LBP	
Trunk Kinematic	mean ± SD	mean ± SD	Manova (p.)	mean ± SD	mean ± SD	Manova (p.)
ROM Hip (°)	20.6 ± 6.9	26.6 ± 9.3	0.018*	25.3 ± 9.1	31.8 ± 8.6	0.018*
ROM Spine (°)	9.9 ± 5.3	7.9 ± 3.5		13.3 ± 7.6	9.2 ± 5.3	
Dynamic stability	mean ± SD	mean ± SD	Manova (p.)	mean ± SD	mean ± SD	Manova (p.)
ROM Chair (°)	22.4 ± 6.1	22.4 ± 7.2	0.461	28.4 ± 6.3	27.5 ± 6.0	0.864
Balancing error (°)	1.8 ± 1.6	2.6 ± 2.5		1.9 ± 2.0	2.3 ± 1.7	
Length of trial (s)	4.3 ± 1.4	4.5 ± 2.1		4.6 ± 2.0	4.3 ± 1.6	
Model Parameters	mean ± SD	mean ± SD	T-Test (p.)	mean ± SD	mean ± SD	T-Test (p.)
Damping ratio	0.3 ± 0.1	0.2 ± 0.1	0.964	0.3 ± 0.1	0.3 ± 0.1	0.775
Natural frequency (rad/s)	3.5 ± 0.9	3.5 ± 0.9	0.940	3.5 ± 1.1	3.8 ± 0.9	0.383
Spine/Hip ratio	mean ± SD	mean ± SD	T-Test (p.)	mean ± SD	mean ± SD	T-Test (p.)
1 st peak ratio	0.47 ± 0.39	0.24 ± 0.16	0.013*	0.58 ± 0.67	0.26 ± 0.29	0.046*
2 nd peak ratio	0.41 ± 0.35	0.18 ± 0.14	0.007*	0.53 ± 0.63	0.31 ± 0.32	0.165

*indicates significant differences (p-value<0.05)

Chair ROM, the time required to regain balance and balancing error (Table 5-2) were not statistically different between LBP and healthy subjects (p>0.05). In addition, the parameters that described the chair angle patterns (damping ratio and natural frequency, Table 5-2) were not significantly different between the two groups.

High correlation was found between the experimental chair angles measured with the motion sensor and those predicted by the model ($R^2 > 0.80$ for all subjects), which demonstrated the goodness of the model proposed. The model parameters predicted in the three repeated trials had ICC of 0.95-0.97, showing that the derived data were highly repeatability.

Lumbar lordosis angle significantly decreased ($p < 0.05$) in LBP subjects, while no differences were found in thoracic kyphosis angle (Figure 5-4). This change in lordosis angle did not correlate with the time required to regain balance, the balance error and the model parameters ($p > 0.05$).

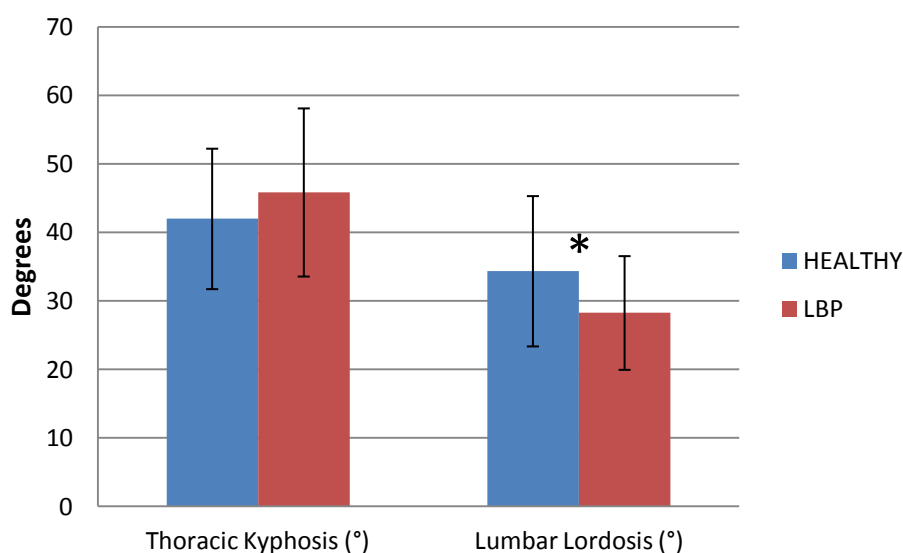


Figure 5-4: The thoracic kyphosis and lumbar lordosis of the two subject groups; *indicates significant difference

5.4 Discussion

The changes in trunk kinematics in LBP subjects might be compensatory strategies to reduce pain and risk of further damage (Shum et al., 2005a, Shum et al., 2005b). Reduced spinal motion might be achieved with increased trunk muscular co-activation, which, as reported in other studies, was found in LBP subjects (Radebold et al., 2000,

van Dieen et al., 2003). The loss in spinal motion was compensated by increased hip joint motion. These results showed to be in agreement with previous studies, where altered mobility of the lumbar spine and hip joint were found (Shum et al., 2005a, Shum et al., 2005b).

LBP subjects effectively employed the compensatory strategies, that is, the alteration in relative spine and hip contributions, to successfully regain the balance on the unstable chair. There were no differences in the damping and natural frequency characteristics between the subject groups. These characteristics determine the shape the chair angle-time curve. This suggests that LBP did not affect the manner in which the dynamic balance was achieved. In addition, the ability to regain balance was not related to the decreased lumbar lordosis in LBP subjects as observed in this study. However, it should be noted that the above conclusion only refer to a medium perturbation such as 20° tilt, and the dynamic response and control may be different in very large perturbation.

It could also be argued that the changes in kinematics of the spine and hip are a cause of back pain, rather than compensation as a result of back pain. The body is adapted to such kinematic changes so that dynamic balance is not affected. This is a limitation of cross-section study which only identifies the relationship between the variables but not the cause and effect of the relationship. Future studies should examine the cause and effect as it would have significant implications on clinical management.

For example, if pain is the cause of kinematic changes, treatment should be directed at pain relief, whereas if pain is the consequence, then rehabilitation programmes should be aimed at restoring the kinematic changes.

Several studies showed that fear avoidance beliefs can be related to chronic LBP, resulting in increased disability (Asmundson et al., 1997, Grotle et al., 2004, Severeijns

et al., 2001, Waddell et al., 1993) and persistent symptoms in subjects with acute LBP (Burton et al., 1995, Fritz et al., 2001, Grotle et al., 2004).

This study provides the evidence that LBP subjects have similar balance performance as healthy subjects without worsening of symptoms and their compensations were effective. These findings should be made known to the patients and this may ease the fear of movement. Clinicians should encourage patients to remain active rather than avoiding movements while they are experiencing LBP.

The present study evaluated motion mostly in the sagittal plane because the chair motion was confined in one direction and the trunk motion on the other plane (transverse and frontal) was negligible. Future studies may change the design of the swinging chair to allow it to move in all directions and examines if trunk motion increases in transverse and frontal planes when the chair is not constrained. In the present study the maximum perturbation applied was only 20°. It may also be useful to impose greater challenges in future work to test if postural stability would be compromised with further increases in the amount of perturbation, but this may be practically and ethically difficult.

5.5 Conclusions

The present study examined dynamic stability and trunk kinematic alteration in LBP subjects during unstable sitting. LBP subjects showed decreased lumbar lordosis angle and reduced spine motions. This was accompanied by increases in hip motions. However, dynamic stability was not compromised by LBP, not only in terms of accuracy but also the manner of regaining balance, indicating the effectiveness of the kinematic strategies performed by LBP subjects. It is suggested that the present findings may help ease the fear of movement of LBP subjects as the symptoms do not seem to affect their balance control.

6 THE ROLE OF TRUNK MUSCLE IN SITTING BALANCE CONTROL IN PEOPLE WITH LOW BACK PAIN

6.1 Introduction

It has been suggested that the main factors contributing to the stability of the trunk are the intrinsic passive stiffness of its structures and the active contraction of the muscles and that these factors are modulated by the neural system (Crisco and Panjabi, 1991, Gardner-Morse and Stokes, 2001, Panjabi, 1992). Moreover, co-contractions of trunk muscles are present to stabilise the spine during several activities (Granata and Wilson, 2001, Cholewicki et al., 1997). Panjabi (1992) proposed that an alteration of the passive structures may be related to a decrease in the intrinsic stiffness that can then lead to increased muscular activity as a compensatory response in order to sustain the stability of the trunk. This was confirmed by several studies which revealed evidence of increased activities of the trunk muscles due to low back pain (LBP) (Fischer and Chang, 1985, Pirouzi et al., 2006). Increased muscle co-contractions (Granata and Marras, 1995b, Marras and Davis, 2001, Marras et al., 2004, Radebold et al., 2000) have also been observed, and may be related to increased spinal stress (Granata and Marras, 1995b, Marras et al., 2004) that may lead to injuries and spinal degeneration (Adams et al., 1996, Gallagher et al., 2005). Moreover, Shum, Crosbie et al. (2007b) studied the effects of back pain on the hip and spine moments during sit-to-stand. They showed that although the moment at the lumbar spine was decreased in the sagittal plane in LBP subjects, moments in transverse and frontal planes were increased. This was believed to be a compensatory strategy to reduce pain.

The control of dynamic balance of the trunk is not fully understood. In comparison with the standing model used in previous studies (Cholewicki et al., 1997, Pirouzi et al., 2006, Sihvonen et al., 1991, Silfies et al., 2005), the present study employed a sitting position to study trunk balance as it would allow us to remove the influence of the legs and to study the role of the trunk in isolation. In particular, the nature of the muscle contraction employed to counterbalance the external moment has not been established. Inverse dynamic analysis has been used to determine if the trunk muscles were generating a flexor or extensor moment to counterbalance the external moment, and the analysis of muscle powers permitted us to evaluate whether the muscular contraction is eccentric or concentric. However, the inverse dynamic model does not allow us to study co-contraction of muscles which may not present during the task (Granata and Marras, 1995b, Marras et al., 2001, Marras et al., 2004, Radebold et al., 2000). This limitation can be addressed by collecting EMG data from trunk muscles, to evaluate any possible co-contraction and to better understand how various muscles contribute to the muscle moment and power.

The purpose of this study was to examine the muscular activities and kinetics of the trunk during unstable sitting and to determine differences in these responses between healthy and low back pain (LBP) subjects. It was hypothesised that there would be differences between LBP and healthy subjects in moment and power distribution between hip and spine joints and in trunk muscles reaction times and co-contraction duration.

6.2 Methods

6.2.1 Subjects

(This paragraph is similar to the paragraph 4.2.1, it has been left in this section to preserve the integrity of the article as it was sent to publication)

Thirty healthy subjects without history of LBP, by self-report and twenty-four subjects with sub-acute (>6 weeks) LBP (Savigny et al., 2009) were recruited for the study. Subjects' characteristics are summarised in Table 6-1 and they were found not to be significantly different between LBP and healthy subjects ($p>0.05$). Exclusion criteria for all subjects were the presence of ankylosing spondylitis, fractures/dislocations of the spine or hips, history of spinal or hip surgery, pregnancy, neurological disorders, cancer and osteoporosis. A visual analogue scale (VAS) was used to record the perceived severity of pain experienced by LBP and the functional ability evaluated by Oswestry Disability Questionnaire (Fairbank and Pynsent, 2000) (Appendix A1). LBP subjects were also asked to indicate where the pain was located (bilateral, left or right side).

Table 6-1 Participants' characteristics (no significant differences between groups)

	Healthy subjects (n=30) mean \pm SD	LBP subjects (n=24) mean \pm SD
Age (yr)	31.73 \pm 8.10	36.83 \pm 11.56
Height (m)	1.676 \pm 0.980	1.689 \pm 0.840
Mass(kg)	63.89 \pm 13.33	68.96 \pm 11.64
BMI (kg/m ²)	22.53 \pm 2.67	24.09 \pm 3.09
Oswestry score (scale 0-100)	N/A	19.83 \pm 8.94
VAS score (scale 0-10)	N/A	3.80 \pm 1.02

N/A=not applicable

6.2.2 Equipment

A custom-made chair (Figure 6-1) was built which was restricted to swing in the sagittal plane. It had feet and leg support to restrict the knee and ankle to a 90° angle, and adjustable belts to restrict lower limb movements. The base of the chair was mounted onto a force platform (Type 9281B, KistlerTM). The chair was built from wood, as metal would interfere with the electromagnetic field generated by the transmitter of the motion tracking system. The swinging mechanism was provided by two low friction ball bearings. Mechanical stops prevented the chair from tilting more than 20° backward and forward.

The movement of the lumbar spine was measured using a three-dimensional motion track system (3SPACE FASTRAK®, Polhemus Inc.) recording at 40 Hz. Two sensors were placed on the subjects' back, one at the sacrum level and one at the first lumbar vertebral level. One further sensor was placed on the chair to track its rotation, which was also used to define the rotation of the lower limbs. Integral dry reusable electromyographic (EMG) electrodes (Biometrics Ltd., type Nos. SX230) were connected to the DataLINK system (DLK900, Biometrics Ltd., Gwent, UK) to record the electrical activities of the paraspinal muscles at a sampling rate of 1000 Hz. The diameter of each electrode was 1 cm and interelectrode distance was 2 cm. The EMG signals were amplified using a single differential amplifier with an input impedance of 107 MΩ, a common mode rejection ratio of 110 dB, and a gain of 1000. After skin preparation with alcohol and shaving of hair, the surface electrodes were secured with double-sided tape bilaterally on the erector spinae, rectus abdominus, external and internal oblique, according with Thomas and Lee (2000). Reference electrode was positioned over the left medial malleolus process.



Figure 6-1: Subject on the custom-made swinging chair.

Reference electrode was positioned over the left medial malleolus process. The chair was placed over the force platform recording at 150 Hz, in order to determine the loads that acted on the system formed by the chair and the subject. Measure Foundry (Data Translation Inc.) software was used to synchronize data acquisition of the devices and to integrate motion and force data. MATLAB® (R2007b, MathWorks Inc.) was used for

data re-sampling and analysis and SPSS (SPSS: An IBM Company) was used for statistical analysis.

6.2.3 Protocol

Subjects were strapped to the chair with the lower limbs and pelvis immobilised, and they were asked to fold their arms across the chest facing forward. The height of the feet support was adjusted to allow the subject to sit in a comfortable position (Figure 6-1). Initially, the subject was tilted twice backwards and forwards by the researcher in a controlled manner to show the range of motion (ROM) of the chair, and then the balanced position (with the chair parallel to the floor) was shown to the subject. Thereafter, the researcher tilted the chair into an angle between 0-20° and following release, the subject was asked to return to the balanced position and maintain it for 5s. This familiarization protocol ended when the subject was able to find the balanced position and to hold it for at least 5 seconds for three repetitions. All subjects were able to complete the familiarization protocol. After the familiarization protocol, the chair was tilted 20°. The chair was released without warning and they were asked to achieve the steady balanced position. Each subject had three attempts to reach the balanced position. The trial was considered successful when the subject was able to reach a steady balanced position without engaging the mechanical stops and maintaining it for 1 second with a maximal oscillation of $\pm 1^\circ$.

6.2.4 Analysis

The muscle moments represented the net effects of the muscles to balance the external moments acting on the hip and lumbar spine. In this manner when external moment was tending to flex, the muscle moment was extensor and vice versa.

Inverse dynamic equations were used to derive the muscle moment acting at the hip and lumbar spine joints using data from the force platform and the motion tracking system (Figure 6-2). Hip joint centre was defined as the caudal endpoint of the trunk defined by Zatsiorsky (2002) and located using digitised landmark points. Centre of mass and inertial properties were calculated using the nonlinear regression equation from Zatsiorsky (2002). The following equation was used to derive muscle moment at the hip M_H and that at the lumbar spine M_S :

Equation 6-1

$$M_H + I_H \varepsilon_S + F_{Rz} d_1 + F_{Ry} d_2 - m_L g d_3 = 0$$

Equation 6-2

$$M_S - I_P * \varepsilon_P - M_H + F_{Hy} * d_5 + F_{Hz} * d_4 - m_H * g * d_6 = 0$$

where I_H and I_P represent the inertia of the lower limbs and the pelvis; ε_S and ε_P are angular accelerations of hip and pelvis; m_L and m_H are the mass of the lower limbs and pelvis; R , H and S represent respectively the position of the centre of rotation of the swinging chair, hip joint centre and L5-S1 joint centre; cm_{LOW} is the centre of mass of the swinging chair cm_{PEL} is the centre of mass of the pelvis; F_{Ry} , F_{Rz} are forces at the centre of rotation of the swinging chair and F_{Hy} , F_{Hz} , are forces components at the hip joint center and d_1 , d_2 , d_3 , d_4 , d_5 , d_6 are distances shown in Figure 6-2 and g gravitational acceleration. Joint velocity was obtained from differentiation of the displacement data acquired by the Fastrak machine, using the 5-point central difference method (Robertson et al., 2013). Muscle power was then calculated by multiplying the muscle moment and the corresponding joint angular velocity.

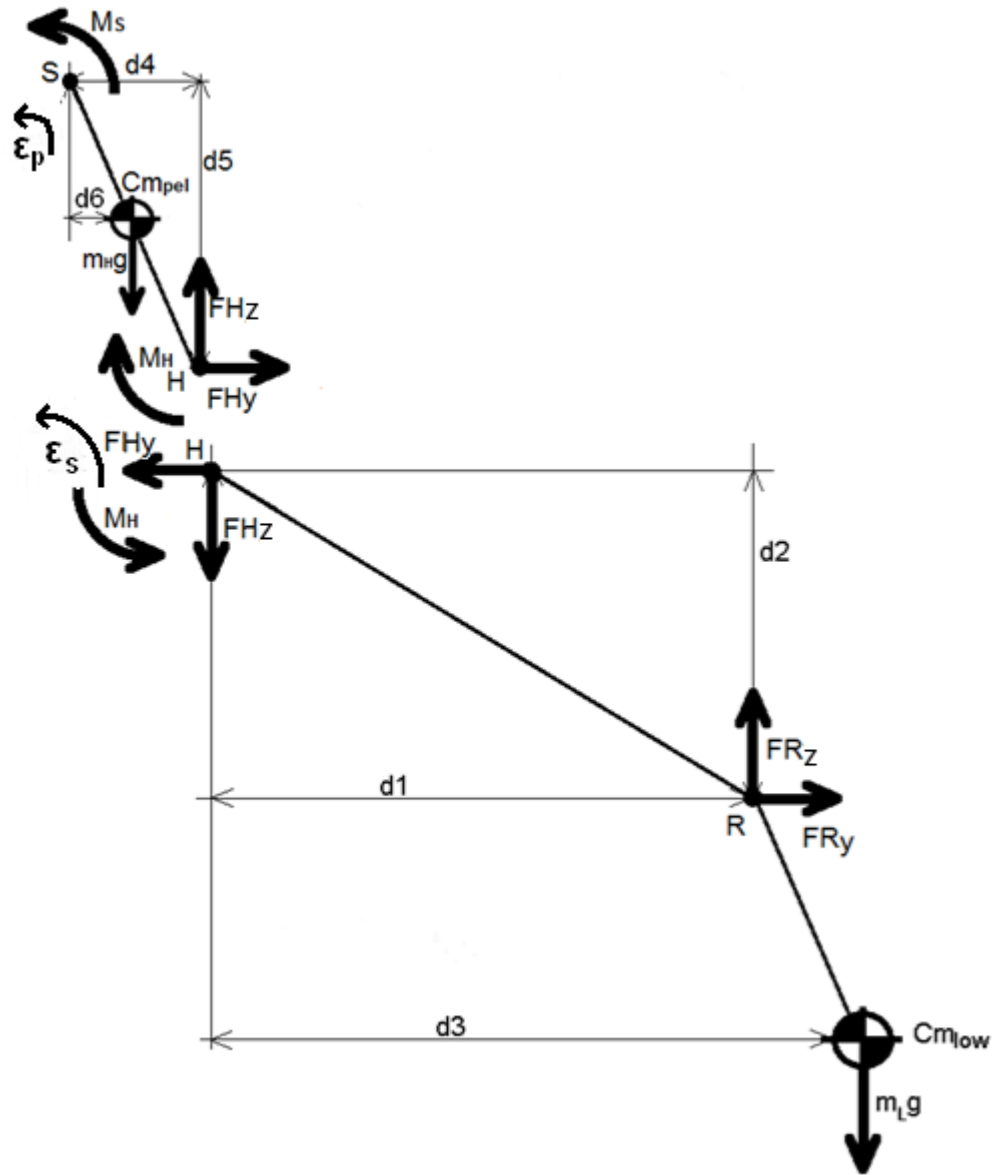


Figure 6-2: Free body diagram for the calculation of the moment at the hip joint and at the lumbar spine joint.

Muscle moment and power data were analysed together to understand the nature of the lumbar spine muscles contractions. When lumbar spine flexor moment was present, a positive power indicated concentric contraction of the lumbar spine flexors while a negative power indicated an eccentric contraction. Similar interpretation could be made

to the lumbar spine extensor behaviour in presence of extensor moment. The role of the muscles in contributing to the moment and power was further studied by EMG.

Anthropometric measures were used to normalize peak to peak power and moment according with Hof (1996) in the following manner:

$$\text{Normalised moment} = \frac{M}{mgl} ; \text{Normalised power} = \frac{P}{m g^{1/2} l^{3/2}}$$

where m=body mass (kg); g = gravity acceleration (9.81 m/s²); l= leg length (m).

EMG data were rectified and band-pass filter (5th order butterworth filter 10-500 Hz) and the linear envelope obtained by applying a low pass filter (5th order butterworth filter with 4 Hz cut-off).

The amplitude of EMG data was normalised using the mean dynamic activity method (Yang and Winter, 1984), whereby the mean of the linear envelope was calculated and considered as 100% of the amplitude.

To detect the on-set and the off-set of each muscle, the method of Stokes et al. (2000) was used. A threshold was calculated as the sum of the mean of the EMG data recorded while the subject was resting plus 3 standard deviation (SD) of that mean; the onset of EMG activity was considered to be the time when the signal exceeded this threshold for at least 150 ms. The off-set time was detected using the same threshold but analysing the EMG signal from the end of the trial till the starting point. Muscular activity was quantified using the total on-set time of the muscle during the trial and expressed as percentage of the total time of the trial. In order to better understand the behaviour of the muscles the percentage of subjects exhibiting total co-contraction of all the muscles at all time instants was also considered.

Muscle reaction time was defined as the time duration between the starting point of the experiment (defined as the point where the chair angle deviated more than 3 SD from

the mean of the chair angle calculated for 0.2 s time window prior to the release) and the on-set of the muscular activity.

6.2.5 Statistics

Multivariate ANOVA was used to evaluate differences between LBP and healthy subjects in the joint velocity, moment and power; the muscle reaction time and the duration of total co-contraction. In order to study any asymmetry in muscle contraction, cross-correlation was performed between left and right sides for internal oblique, external oblique, rectus abdominis and erector spinae for each subject. Gender was found to have no effect on any of the dependent variables ($p > 0.05$) and so male and female data were combined together for statistical analysis between each subject group. For all statistics significant level was set at 0.05.

6.3 Results

Hip angular velocity variables were found to be significantly increased in LBP subjects ($p < 0.05$), (Table 6-2). Lumbar spine joint velocity tended to be decreased for the LBP subjects, but such difference was not statistically significant ($p > 0.05$)(Table 6-2).

There were no significant differences between LBP and healthy subjects regarding the normalised hip and lumbar spine powers and muscle moments (Table 6-2) ($p > 0.05$).

Figure 6-3 illustrates the typical kinetic and EMG responses of a LBP subject, after releasing the chair. The moment was positive throughout the trial, indicating that the muscles were producing a net flexor moment. All muscles contracted together, indicating co-contraction.

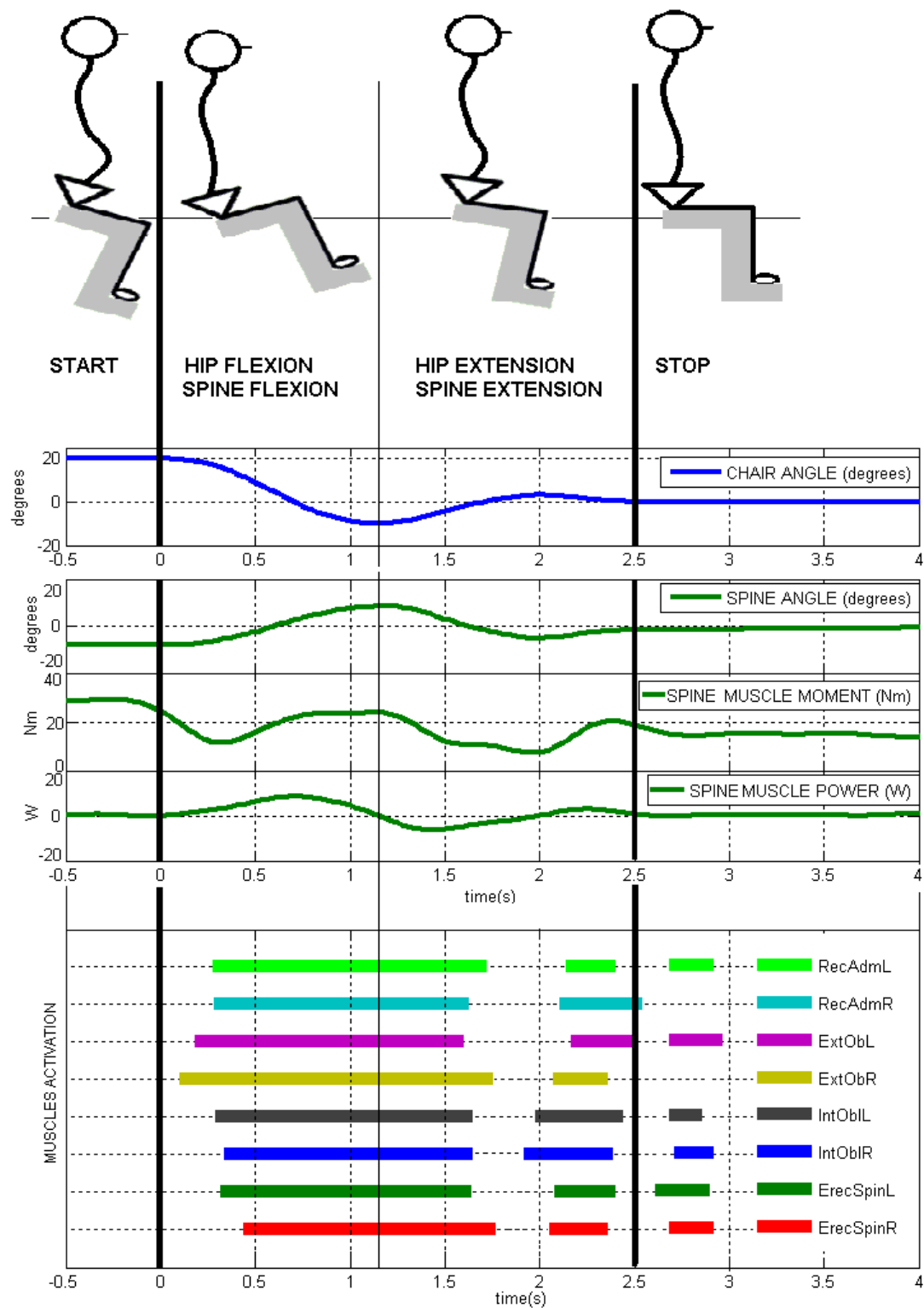


Figure 6-3: Motion of the lumbar spine and EMG activity during the different phases of the balancing task for one LBP subject. The top graph shows the angular displacement of the swinging chair during the experiment, starting from 20° and stopping at 0°. Angular displacement, muscle moment, muscle power pattern of the lumbar spine joint and trunk muscles onset time are shown.

Table 6-2 Mean velocity, moment and power data for hip and lumbar spine joints

	Hip joint		Lumbar spine joint	
	Healthy mean \pm SD	LBP mean \pm SD	Healthy mean \pm SD	LBP mean \pm SD
Velocity RMS (rad/s)	0.35* \pm 0.11	0.44* \pm 0.11	0.16 \pm 0.09	0.13 \pm 0.07
Velocity P-P (rad/s)	1.50* \pm 0.53	1.91* \pm 0.56	0.71 \pm 0.36	0.58 \pm 0.34
Normalized Moment	0.083 \pm 0.04	0.081 \pm 0.03	0.084 \pm 0.05	0.089 \pm 0.03
Normalized Maximum Moment	0.087 \pm 0.03	0.078 \pm 0.02	0.072 \pm 0.03	0.067 \pm 0.02
Normalized RMS Power	0.06 \pm 0.04	0.06 \pm 0.03	0.013 \pm 0.01	0.010 \pm 0.01
Normalized P-P Power	0.20 \pm 0.14	0.19 \pm 0.11	0.06 \pm 0.05	0.04 \pm 0.04

*indicates significant differences (p -value<0.005)

The power remained positive from the beginning of the trial until 1.1s, indicating that during the co-contraction, concentric flexor power was stronger than the eccentric extensor power, resulting in a net positive power; from 1.1s until 1.7s the power became negative and the pattern reversed, with concentric extensors power stronger than the eccentric flexor power. Before reaching the balanced position the power became positive again. As shown in Figure 6-3, muscular contractions started soon after chair release, and this was similarly observed in all subjects. The onset time of all muscles was significantly reduced for the LBP compared to the healthy subjects (p <0.05) (Figure 6-4). Furthermore, the percentage duration of contractions of all muscles was longer for LBP subjects during the trial compared with healthy subjects (p <0.05) (Figure 6-5).

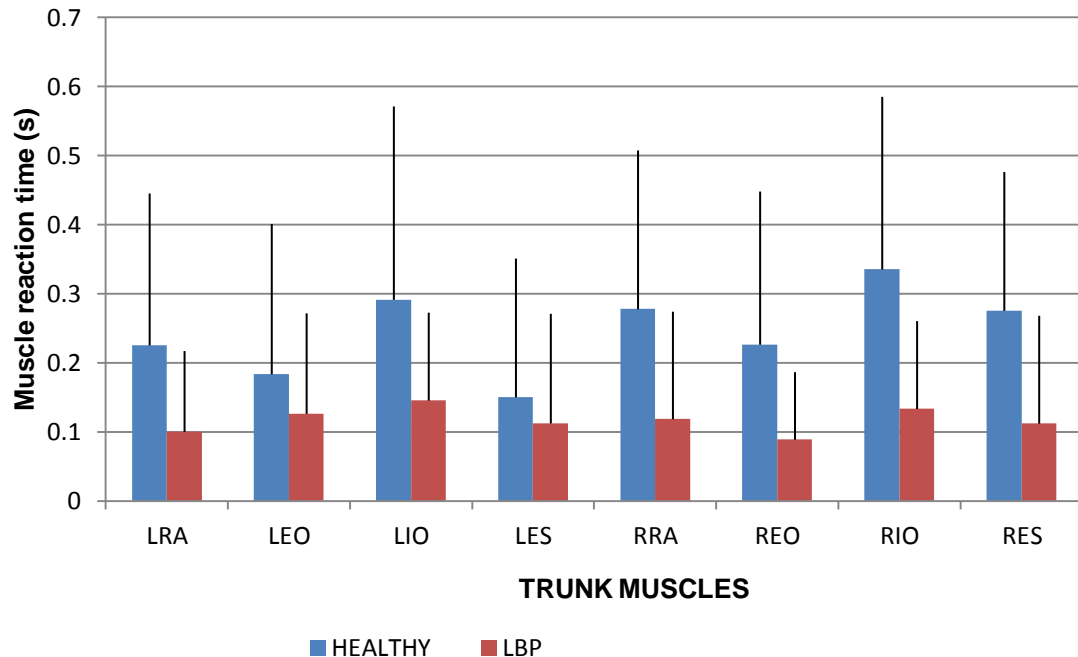


Figure 6-4: Reaction time of trunk muscles during unstable sitting. Significant differences were present between the two groups ($p < 0.05$). (L=left, R=right, Rectus Abd = Rectus Abdominus, Ext O = External Oblique, Int O = Internal Oblique, Erector S = Erector Spinae).

The percentage of LBP subjects which showed total muscle co-contraction was increased when compared to healthy subjects, for most of the time instants as shown in Figure 6-6. The mean percentage of all time instants was significantly higher for LBP subjects ($p < 0.05$) (12.6% for healthy and 35.2% for LBP subjects). Mean cross-correlation coefficient between left and right side EMG signals was significantly decreased in LBP subjects for the erector spinae muscle (healthy: 0.671, LBP: 0.557, $p < 0.05$). No significant differences in the mean of the cross correlation were found for the other muscles ($P > 0.05$).

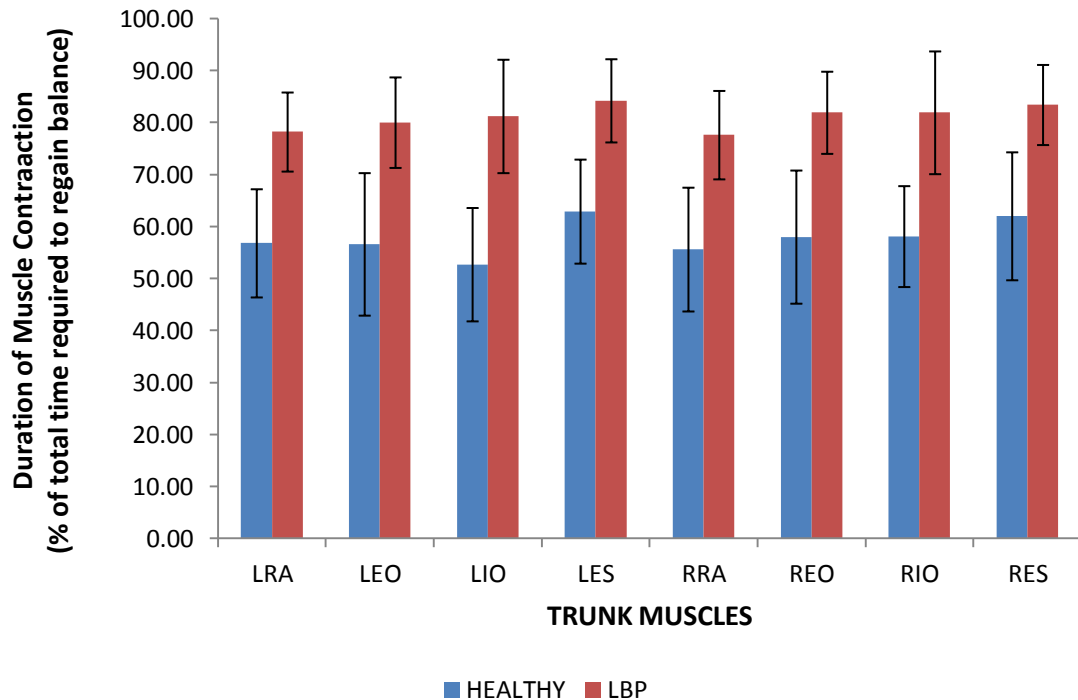


Figure 6-5: Duration of muscle contraction as percentage of total time required to regain balance. Significant differences were present between the LBP and healthy subjects ($p < 0.05$).

6.4 Discussion

During the balancing task, trunk muscles generated hip and lumbar spine moments in order to counterbalance the external moments and to permit to the subjects to regain the balanced position. The results showed that moments generated by LBP subjects were similar in pattern and magnitude to those generated by healthy subjects. These are in contrast with findings of other experiments where subjects were performing different tasks (Shum et al., 2007b) where alterations in the kinetic pattern were found. As shown by Figure 6-3, muscle moments oscillated until the balancing position was reached. The joint powers gave information about the nature of the muscle contraction, distinguishing between eccentric and concentric contraction as described above. Similar to the muscle moments, muscle power data were not significantly different in

magnitude between the two groups of subjects. From a kinetic point of view, there were no alterations due to LBP, but EMG results showed that muscular contraction time for each trunk muscle was increased in both trials for LBP subjects (Figure 6-5). The decrease in muscle reaction time observed in this study does not agree with the results of previous research (Radebold et al., 2001) , which reported the opposite. This could be due to the differences in activities being examined or how the experiment was conducted. In this study, the subjects were given the opportunity to familiarize themselves with the swing before the test.

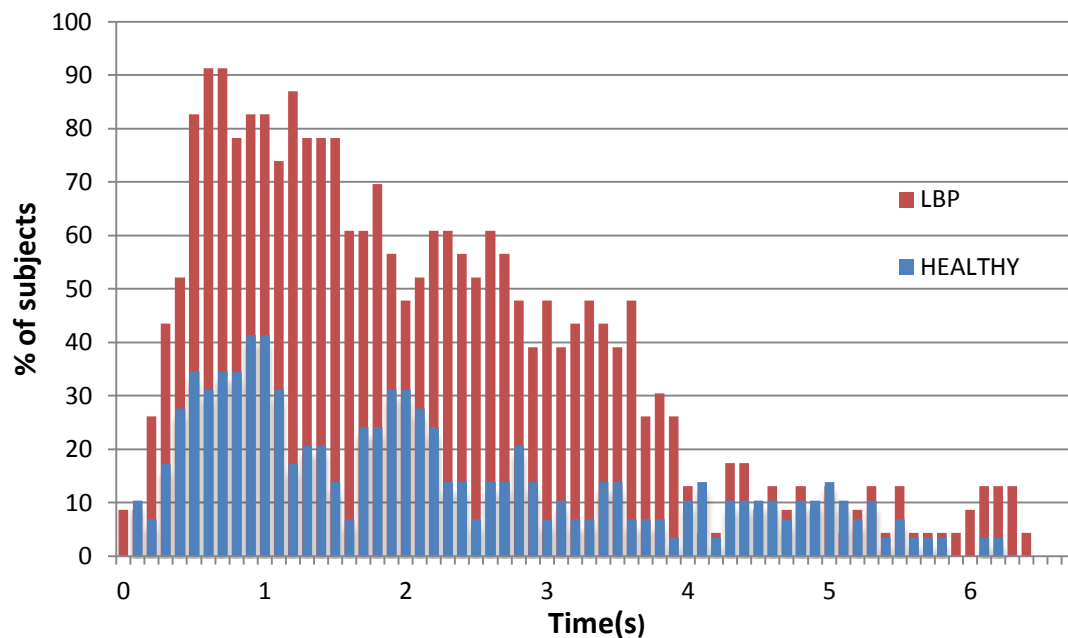


Figure 6-6: On-off total muscular co-contraction as percentage of number of LBP and healthy subjects.

In fact, more than 50% of the LBP subjects in our study had some of the muscles already activated before the start of the trial, suggesting that LBP subjects were preparing for the swing. There were increased number of people with co-contraction of muscles among the LBP subjects, indicating that the agonist and antagonist muscles were actively working together to balance the external moment. These results were in

accordance with previous studies which also found increased trunk muscular activity in LBP subjects during various activities (Cholewicki et al., 1997, Sihvonen et al., 1991, Silfies et al., 2005), and responding to an external perturbation in a semi-seated position (Hodges et al., 2009). It is possible that the changes in muscle activity may be a compensatory strategy to account for the decrease in the intrinsic stiffness due to LBP (Panjabi, 1992). It may be a protective response to pain in order to limit spine motion and reduce the risks of further damage to the spine tissue (Fischer and Chang, 1985). However, it is also possible that the changes in EMG may be the cause of back pain rather than a response to pain. The EMG response in LBP subjects could be less efficient than those in healthy subjects because more muscular activity and increased co-contraction is required to balance external moments. The prolonged increase in muscular activity may then lead to increasing spinal stress (Marras et al., 2001, Marras et al., 2004, Granata and Marras, 1995b), muscle fatigue and contracture (Thomas and Lee, 2000), resulting in pain. As this is a cross-sectional study, although we are able to examine the effect of pain on muscular responses, the cause and effect of the association cannot be ascertained, and this needs to be further investigated in future study. Clinically, if it is shown that the changes in EMG pattern is a cause of LBP, it would be useful to explore the feasibility of using the swinging chair as a biofeedback training tool for LBP patients to restore the motor control of the trunk muscles. The current experimental set up may also be provided with a visual tool that can be used to show a target that the subject should achieve in trunk control. Combination of the visual training tool with EMG of trunk muscles can provide useful information about muscles coordination and may be used to evaluate the progress of rehabilitation programme.

On the other hand, if the changes in EMG are shown to be a consequence of pain, the above biofeedback training may not be helpful. Therapeutic intervention should be directed to relieve pain so that the EMG patterns could be restored, avoiding continued

increase in spinal stress and aggravation of the symptoms (Marras et al., 2001, Marras et al., 2004, Granata and Marras, 1995b). However, it is possible that pain and EMG changes are operating in a viscous circle, reinforcing each other leading to a chronic condition. Therapy should help break this viscous circle, with pain relief to reduce symptoms together with exercise to restore muscular function.

Moreover a number of studies established that fear avoidance beliefs resulted in increased disability and persistent symptoms in LBP subjects (Grotle et al., 2004, Severeijns et al., 2001). This study demonstrates that LBP subjects were able to perform the balance tasks with similar kinetic profile as the healthy subjects and the ability to perform the task was not compromised, despite the changes in muscle response. This may suggest to clinicians to encourage patients to remain active rather than to avoid movements.

6.5 Conclusions

The present study shows how trunk muscles generated an internal moment to counterbalance the external perturbation and regain balance during unstable sitting. There were altered muscle contraction patterns and significant increases in the percentage of LBP subjects in co-contraction. These active mechanisms appear to be effective in maintaining the stability of the spine, as trunk balance was not compromised and moment distribution was not altered in symptomatic subjects. This study clearly shows the association between LBP and changes in muscle responses, but the cause and effect of this association is still uncertain. Longitudinal studies should be carried out in the future to examine this association; as such understanding will have significant implications on treatment planning. Future research should also look at the effects of more complicated tasks (e.g. random three-dimensional sitting perturbation) on the muscle response of the lumbar spine.

7 GENERAL DISCUSSION

7.1 Overall discussion of the results

This series of experimental studies evaluated the possible alterations in biomechanical variables including muscular functions, viscoelastic properties, kinematic and kinetic of the trunk in non-specific low back pain sufferers. This was achieved by designing an experimental task where subjects attempted to regain the balance using trunk motion on a custom-made swinging chair after it was subjected to an unexpected tilt. As it was shown by literature review, different studies evaluated biomechanical response to a certain activity in LBP subjects, but in some cases the motion was not limited to the trunk, which included biases from the lower limbs, results were contrasting and the relationship between different biomechanical variables evaluated was unclear. This study was designed to evaluate biomechanical variables to better understand mechanisms underlying back pain and the relationship between mechanical properties, muscle functioning, motion, loading strategies and performance of the trunk.

The study was designed by taking into account the limitations found in previous studies investigating trunk biomechanical behaviour. Sitting position was used to remove influence of the lower limbs that may have contributed to the task (Moorhouse and Granata, 2005) and considering passive and active components together (Nichols and Houk, 1976). Furthermore this position was chosen together with an active balancing task to overcome the lack of ecological validity as in some studies subjects were prone or semi-upright sitting i.e., the effect of gravity was altered, or there was no dynamic response to a perturbation (Brown and McGill, 2009, Gardner-Morse and Stokes, 2001, Hodges et al., 2009, Latimer et al., 1996c, McGill et al., 1994), Comparing with a previous study (Hodges et al., 2009), viscoelastic properties were derived combining trunk moment and motion data in a second order model, but in the model proposed in

this study inertial effects were also included to improve the estimation of mechanical properties. The goodness of the model used was shown by the high correlation and small RMSE between moment calculated using the second order model and the moment calculated with inverse dynamic equations as it is shown in chapter 3.

This study evaluated the overall mechanical properties of the trunk by taking into account of both the active and passive trunk structures, in contrast with several studies evaluating mostly mechanical properties of passive structures (Colloca and Keller, 2001, Keller et al., 1987, Latimer et al., 1996c, Shum et al., 2013, Tafazzoli and Lamontagne, 1996). A dynamic task was chosen because, as mentioned in the literature review, previous research showed that viscoelastic properties of the trunk were largely modulated by muscle activation and to a smaller extent the passive moment generated by the joints (Brown and McGill, 2009, Cholewicki et al., 2000a, Colloca and Keller, 2001).

Dynamic stability was evaluated using a different approach compared to earlier work by using centre of pressure sway (Cholewicki et al., 2000a, Preuss et al., 2005, van Dieen et al., 2010). A more detailed analysis was performed, modelling the response to the perturbation with a second order equation (Bajd and Vodovnik, 1984, Ditroilo et al., 2011a) in order to characterise the damping and natural frequency characteristics of the response. In addition duration and accuracy of the balancing task were measured to evaluate subject's performance.

Activities of trunk muscles were recorded and analysed with surface EMG; inverse dynamic approach was used to determine the nature of the moment generated by the muscle (i.e. flexor/extensor) that counterbalance the external moment, while muscle power analysis was used to determine the nature of the muscle contraction (i.e. eccentric or concentric). This model could be used when no co-contraction of muscles were present during the task, but this could not be assumed for this study (Granata and

Marras, 1995b, Marras et al., 2001, Marras et al., 2004, Radebold et al., 2000). EMG data from trunk muscles were collected (Radebold et al., 2000) to overcome this limitation and to evaluate muscular co-contraction and how various muscles contributed to the muscle moment and power.

The study showed that all subjects responded in a similar manner to the perturbation, in particular subjects flexed and extended hip and spine to control movement of the swinging chair until reaching balanced position. The motion of the chair, which was used to explain how the chair was controlled by subjects, followed a damper oscillator pattern in both subject groups. Furthermore the response of the trunk to the perturbation proposed in this study was mainly elastic as the damping coefficient calculated in chapter 3 was almost zero for all subjects, with no distinction between healthy and LBP groups. This may be due to the velocity of the perturbation, which permitted subjects to respond to a change of the external moment with a sudden proportional change of trunk motion. This result was in agreement with a study found previously (Brown and McGill, 2009), whereby trunk response showed to be mainly elastic, with damping coefficient close to zero. In previous vitro studies, damping was observed only in quite big changes in velocity (Crisco et al., 2007).

As it is described in chapter 4, there were no differences in response characteristics (damping ratio and natural frequency) and in the ability to regain balance (task duration, balancing error) between LBP and healthy subjects, implying that dynamic stability was not compromised as a result of LBP.

On the other hand, LBP subjects exhibited alterations in trunk kinematics, in particular decreased spinal motion was compensated with increased hip motion, similar to what was shown by other studies evaluating trunk kinematic while performing everyday life activities as putting on a sock, stand-to-sit, sit-to-stand and picking-up (Shum et al., 2005a, Shum et al., 2005b). These alterations may be explained as a different muscular

strategy, as it is shown in chapter 5, where there was an increased activity of trunk muscles, as also found in previous studies (Cholewicki et al., 1997, Fischer and Chang, 1985, Hodges et al., 2009, Sihvonen et al., 1991, Silfies et al., 2005). This increased trunk muscle activity seemed to produce a reduction of spinal motion. As a consequence hip strategy seemed to be used to compensate this loss of spinal movement and not affect dynamic stability.

Kinematics and muscular alterations found in LBP subjects agreed with results in chapter 3 where trunk stiffness coefficient was found increased in symptomatic subjects. Moreover static standing posture of the trunk was evaluated, showing a decrease in lumbar lordosis in LBP subjects that was not correlated with alteration in task performance.

It can be summarised that LBP subjects used a different strategy with decreased spinal motion together with increased hip motion. These kinematic alterations were in agreement with increasing in dynamic trunk stiffness found in this study in LBP subjects. These alterations seems to be achieved by LBP subjects with altering trunk muscular activity, in particular, as it shown in chapter 5, increasing muscles co-contraction time and reducing muscles latencies.

The present study also evaluated kinetics of the trunk during the balancing task and it showed that moment and power produced by trunk muscles were similar in both healthy and LBP subjects groups. As the trunk muscular activity was higher in LBP, this may imply that the trunk response to the balancing task was less efficient than in asymptomatic subjects.

Trunk muscles are involved in spinal stability and motion. The findings of this study suggest that the increase in muscles activities and co-contractions is associated with the increase in stiffness of the spine. This may, in turn, be related to the reduction in motion, the changes in kinematic strategies and elastic properties as observed

elsewhere in this thesis. The findings of the various chapters of the thesis are coherent, allowing firm conclusions to be made about the kinematic, EMG and functional changes of the spine. The kinematic changes may be a mechanism to minimise pain, protect the spine and increase its stability (Fischer and Chang, 1985, Shum et al., 2005a, Shum et al., 2005b). In addition, muscular reaction times results seemed to agree with this theory where alterations may be a protective strategy that LBP subjects adopted to prevent further risks. This is indicated by shorter reaction times of trunk muscles in LBP subjects. In some cases trunk muscles were already contracted before starting the trial, demonstrating that LBP subjects were more concerned about the task and they contracted muscles to prevent dangerous trunk movements.

On the contrary, increasing in trunk stiffness, reducing in spinal motion and increasing in trunk EMG activity may be the cause of the pain rather than a response to the pain. It has been demonstrated by previous studies that increasing in EMG activity in LBP subjects may lead to increasing spinal stress (Granata and Marras, 1995b, Marras et al., 2001, Marras et al., 2004), muscle fatigue and contracture (Thomas and Lee, 2000), resulting in persisting pain.

7.2 Clinical implications

The cause and effect relationship was not established in this thesis as the study was a cross-sectional study, and only relationships between variables were examined. From a clinical point of view if alterations in biomechanical variables were the cause of LBP, it would be useful to explore the feasibility of using the custom made swinging chair as a biofeedback training exercise for people with non-specific LBP to restore the motor control of the trunk muscles and, in turn, to correct trunk kinematic and to reduce trunk stiffness. The training exercise should involve a random perturbation and a visual tool to show a target to be achieved through controlling trunk motion. The training tool should

be also combined with surface EMG to evaluate functioning and coordination of the trunk muscles and to check the progress of rehabilitation programme.

On the contrary, if alterations of EMG, viscoelastic and kinematic trunk variables were a consequence of the pain, treatments should be directed at pain relief, to reduce trunk stiffness, improve spinal kinematic and to decrease spinal stress related to increasing in trunk muscles activity (Granata and Marras, 1995b, Marras et al., 2001, Marras et al., 2004), in order to avoid persisting and aggravation of symptoms.

Nevertheless, pain and biomechanical alterations may be concatenated, reinforcing each other, leading to a chronic condition. If this is the case, therapy should help break this vicious circle, with pain relief to reduce symptoms together with training exercises and rehabilitation to restore correct spinal kinematics, muscular function and to reduce trunk stiffness.

Besides, a number of studies established that fear avoidance beliefs resulted in increased disability (Asmundson et al., 1997, Grotle et al., 2004, Severeijns et al., 2001, Waddell et al., 1993) and persistent symptoms in LBP subjects (Burton et al., 1995, Fritz et al., 2001, Grotle et al., 2004). This study demonstrated that LBP subjects were able to perform the task with similar kinetic profiles as the healthy subjects, without compromising dynamic stability and performance, despite the changes in muscle responses, trunk stiffness and kinematic variables. This may suggest to clinicians to encourage patients to remain active rather than to avoid movements.

7.3 Limitations, improvements and future directions

Some improvements can be suggested as next step forward to understand the mechanism behind non-specific LBP not explained through this research. Firstly an interventional study may be performed to establish cause and effect relationship between pain and altered biomechanical variables in LBP subjects. This may be done

with pain-induced studies, in order to evaluate the direct consequence of the pain on the variables analysed in this study, but this may be ethically and practically difficult to perform. A more practicable way to evaluate this relationship may be a pain relief study, where subjects before and after a pain relief medications may be asked to perform a balancing task, and possible improvements in terms of trunk stiffness, muscles functioning and trunk kinematic may be evaluated.

The model used to evaluate dynamic stability was a second order linear model, and it showed to be highly reliable and valid for the proposed task and testing apparatus. In future studies, design of the testing apparatus may be adapted to deliver different types of perturbation rather than a step input, for example sinusoidal perturbation that will permit to evaluate also responses at different frequency values or a random input that will permit to use a system identification approach to evaluate subject response to the perturbation. In addition a random perturbation will permit to evaluate everyday life activity as driving a car with a better reproducibility to what is happening in real life. The perturbation delivered was relatively small (20°), and so increasing task difficulty may highlight differences in dynamic stability and task performance, in contrast to what was found in the research, but it may increase risks for the subjects. More complex loading conditions may require the development of more sophisticated mathematical models to describe the response of the spine. For instance multidimensional viscoelastic model, which used a combination of mechanical elements (springs and dashpots) may be employed evaluate the viscoelastic properties in different experimental condition. Furthermore an identification system approach may be used to evaluate the compliance of the system by studying the mathematical relation between perturbation inputs and trunk response outputs (Flugge, 1975).

The study demonstrated that LBP subjects increased hip motion to compensate the reduction in lumbar spine movements, it could be interesting to monitor muscles that

are involved in hip motion i.e. gluteus maximus, rectus femoris, tensor fasciae latae, that may have altered activities as a result of no-specific LBP. Moreover it could be also interesting to evaluate if other back muscles (e.g. latissimus dorsi) are influenced by this condition. Kinematic analysis of the trunk can be improved by adding further sensors to track in more detailed manner the motion of different trunk segments (e.g. thorax segment) and to evaluate if compensatory strategy may involve also other trunk segments rather than hip and lumbar spine joints. Psychological tests could be added to evaluate fear and concerns during the experiment that may play a role in the performance of the LBP subjects. Recruiting older people or people with stronger symptoms could be useful to evaluate long term impact of the LBP and how the condition develops. Recruiting people without LBP but with high risk of developing it may be useful to check for possible markers that may help clinicians to predict the onset of the pain. As mentioned above, it could be useful to evaluate the possibility to use this apparatus as a biofeedback training exercise, but before cause and effect relationship needs to be clarified.

A development of this research may involve the design of a trunk motion simulator for in-vitro studies. Data from this study may be used to control a testing apparatus for in-vitro tests in cadaveric specimens of the trunk, similar to what has been done for the knee joint where a custom made apparatus simulated knee joint flexion-extension according with data obtained in in-vivo studies (Ostermeier et al., 2007, Wünschel et al., Zavatsky, 1997). A robotic apparatus, controlled with kinematic or kinetic data derived from this research, may be used to control cadaveric specimens in order to simulate hip and spinal flexion/extension and may be used to evaluate outcomes of different surgical procedures performed in-vitro, before testing them on patients. A future step may be the use of the trunk motion simulator with a specimen reconstructed from a magnetic resonance image of a subject spine, which will allow clinicians to better understand

which interventional procedure is the best option for a certain subject and to evaluate outcomes and cost effectiveness for the proposed procedure before to operate on the subject.

8 CONCLUSIONS

The findings reported in this thesis showed that subjects with non-specific LBP had reduced lumbar spine mobility, but with increased co-contraction of trunk muscles, which in turn was related to increased effective trunk stiffness. This strategy permitted LBP subjects to perform a short task of regaining balance after a small tilt on a swinging chair, without increasing/inducing pain and with the same effectiveness of healthy subjects. Differences were found in the strategy, where reduction in lumbar spine motion was compensated with increased hip motion.

The thesis also showed decreased efficiency as a result of the increased co-contractions for trunk muscles. These findings are shown by literature to be linked to increased spinal stress. As this was a cross-sectional study, the cause and effect relationship was not established, and it can be argued that pain can be the cause of the alterations in biomechanical variables rather than the consequence.

Clinically, if pain was the cause of the biomechanical alterations treatments should target pain, with pain relief medication to reduce symptoms and their progression. On the other hand, if the pain was the consequence, clinicians should suggest training exercise to restore a correct trunk biomechanical behaviour to avoid pain and the custom made swinging chair may be used as a biofeedback training exercise. It may be possible that pain and biomechanical alterations were concatenated, reinforcing each other, leading to chronic conditions. In this case, treatment should break this vicious circle, in particular clinicians should suggest a rehabilitation program that includes both pain relief medication and training exercise to reduce symptoms and restore correct trunk biomechanics.

The thesis has made a significant contribution to our understanding of the dynamic properties of the spine, and how they are related to balance control and muscle

activities. Such understanding is clinically important as the study showed how LBP affects biomechanical variables and their relationship: in particular LBP subjects exhibited a different balancing strategy, increasing muscular activity to reduce spinal motion which leads to an increase in trunk stiffness. It is hope that the work will stimulate further research; in particular, establishing the cause and effect relationship between pain and alterations in biomechanical variables as this was critical for rationalising our management program. This thesis represents a significant step forward in understanding of LBP and related biomechanical mechanisms in people with non-specific pain and hope it may bring some benefits to millions of LBP sufferers.

APPENDICES

A1. Participant's forms and questionnaires



PARTICIPANT CONSENT FORM

Title of Research Project: *Dynamic properties of the lumbar spine in subjects with arthritic conditions and non-specific back pain*

Brief Description of Research Project: This study is an investigation of the physical properties of the spine for people with arthritis and those with non-specific low back pain. Measurements of the movements and electrical activity of the back will be made by electronic sensors and electrodes attached to your body while you sit on a seat designed to move in a controlled pattern in order to mimic everyday life activity such as sitting on a bus. The research will take place in the Biomechanics laboratory, Roehampton University. The data collection will take place in 1 session of no more than 2 hours.

Investigator Contact Details:

Name: Marco Freddolini, PhD student
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Consent Statement:

I have read the Information Sheet and I have understood what this research study involves and I agree to take part in it.

I am aware that I am free to withdraw at any point without fear of prejudice.

I understand that the information I provide will be treated in confidence by the investigator and that my identity will be protected in the publication of any findings. If I decide to withdraw, I understand that my data already collected will be used but all the other data will be destroyed.

I have understood the research named above and I agree to participate.

Name

Signature Date

Please note: if you have a concern about any aspect of your participation or any other queries please raise this with the investigator. If you have a concern about any aspect of your participation or any other queries, please raise this with the investigator. However if you would like to contact an independent party please contact the Dean of School (or if the researcher is a student you can also contact the Director of Studies or co-supervisor of this study). Their contact details are stated as follow:

Dean of School of Human and Life Sciences

Name: Michael Barham
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Director of Study:

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DEBRIEFING FORM

Subject ID Number:

Title of Research Project:

Dynamic properties of the lumbar spine in subjects with arthritic conditions and non-specific back pain

Brief Description of Research Project:

This study was an investigation of the physical properties of the spine in people with arthritis low back pain. Measurements of the movements and electrical activity of the back has been made by electronic sensors and electrodes attached to your body while you were sat on a seat designed to move in a controlled pattern in order to mimic everyday life activity such as sitting on a bus. This study will develop our understanding of the dynamics of the back in those with arthritis and low back pain.

Please feel free to contact the principal investigator with any questions.

Principal Investigator Contact Details:

Name: Marco Freddolini, PhD student
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Address: Roehampton University, Whitelands College, Holybourne Avenue, London, SW15 4JD
E-mail: marco.freddolini@roehampton.ac.uk
Tel: +44 (0)20 8392 3549

Your right to withdraw

You have the right to withdraw from this study at any time without fear of prejudice. If you decide to withdraw, please tell the principal investigator at the earliest possible opportunity. If you withdraw, it may be beneficial to use your data already collected up to the point of withdrawal but all other data will be destroyed. The data that have been collected will only be retrieved and analysed using the subject ID number assigned to your data set, and your anonymity will be maintained at all times.

Data analysis

The data collected will be analysed to evaluate dynamic properties of the back in people with non-specific back pain and arthritic condition in order to better understand the mechanism underlying back pain and as a consequence will enhance clinical work, helping clinician to rationalise and improve treatment for back pain, increasing the benefits to the patients.

Further information and contact details.

You will be given the opportunity to consult a registered physiotherapist, Jonathan Williams who is associated with the research group, after the test if you have experienced pain. There is no charge for this service.

For further information about this research project, please contact Marco Freddolini (Tel: +44 (0)20 8392 3549, Email: marco.freddolini@roehampton.ac.uk).

If you have a concern about any aspect of your participation or any other queries, please raise this with the investigator. However if you would like to contact an independent party please

contact the Dean of School (or if the researcher is a student you can also contact the Director of Studies or co-supervisor of this study). Their contact details are stated as follow:

Dean of School of Human and Life Sciences

Name: Michael Barham
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Finally, thank you again for participation in this research.

Participant information

Please complete this questionnaire

Subject ID:

Date:

Gender:

Age:

Do you have any of the following conditions?

- | | | |
|--------------------------|------------------------------|-----------------------------|
| - Osteoporosis | <input type="checkbox"/> YES | <input type="checkbox"/> NO |
| - Cancer | <input type="checkbox"/> YES | <input type="checkbox"/> NO |
| - Neurological Disorders | <input type="checkbox"/> YES | <input type="checkbox"/> NO |
| - Ankylosing Spondylitis | <input type="checkbox"/> YES | <input type="checkbox"/> NO |
| - Others | Please specify:..... | |

Do you have any medical implant? YES ☐ NO ☐

If yes please specify:.....

Are you pregnant? YES ☐ NO ☐

Have you ever had

- | | | |
|--|------------------------------|-----------------------------|
| - fractures or dislocation of the spine? | <input type="checkbox"/> YES | <input type="checkbox"/> NO |
| - fractures or dislocation of the hips? | <input type="checkbox"/> YES | <input type="checkbox"/> NO |
| - spinal surgery? | <input type="checkbox"/> YES | <input type="checkbox"/> NO |
| - hip surgery? | <input type="checkbox"/> YES | <input type="checkbox"/> NO |

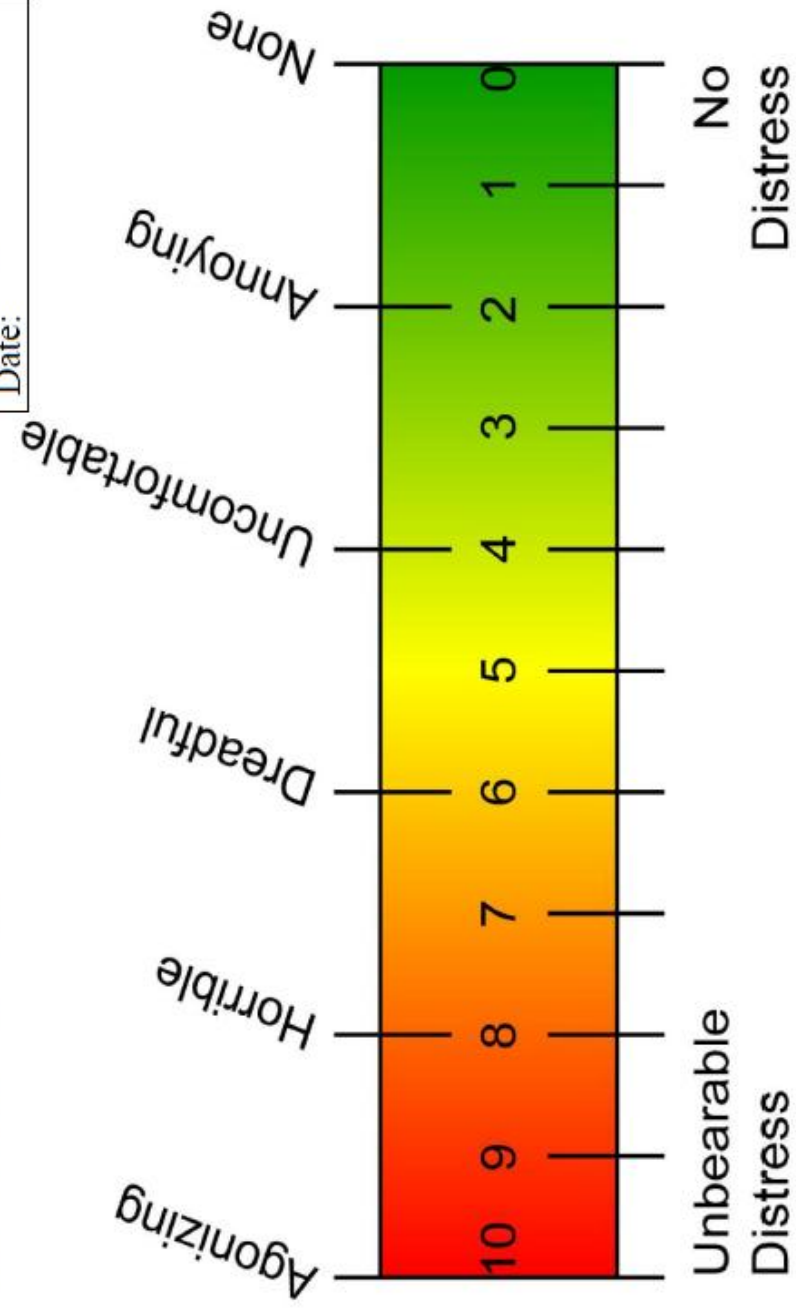
Are you suffering from low back pain? YES ☐ NO ☐

If YES, which is the cause of your pain?

- | | |
|-----------------------|--------------------------|
| - Arthritic condition | <input type="checkbox"/> |
| - Unknown condition | <input type="checkbox"/> |
| - Other | Please specify:..... |

Subject ID:
Date:

SEVERITY OF THE PAIN



SUBJECT ID:.....
DATE:.....

Oswestry Disability Questionnaire

This questionnaire has been designed to give us information as to how your back or leg pain is affecting your ability to manage in everyday life. Please answer by checking **one box in each section** for the statement which best applies to you. We realise you may consider that two or more statements in any one section apply but please just shade out the spot that indicates the statement **which most clearly describes your problem**.

Section 1: Pain Intensity

- ☐ I have no pain at the moment
- ☐ The pain is very mild at the moment
- ☐ The pain is moderate at the moment
- ☐ The pain is fairly severe at the moment
- ☐ The pain is very severe at the moment
- ☐ The pain is the worst imaginable at the moment

Section 2: Personal Care (eg. washing, dressing)

- ☐ I can look after myself normally without causing extra pain
- ☐ I can look after myself normally but it causes extra pain
- ☐ It is painful to look after myself and I am slow and careful
- ☐ I need some help but can manage most of my personal care
- ☐ I need help every day in most aspects of self-care
- ☐ I do not get dressed, wash with difficulty and stay in bed

Section 3: Lifting

- ☐ I can lift heavy weights without extra pain
- ☐ I can lift heavy weights but it gives me extra pain
- ☐ Pain prevents me lifting heavy weights off the floor but I can manage if they are conveniently placed eg. on a table
- ☐ Pain prevents me lifting heavy weights but I can manage light to medium weights if they are conveniently positioned
- ☐ I can only lift very light weights
- ☐ I cannot lift or carry anything

Section 4: Walking*

- ☐ Pain does not prevent me walking any distance
- ☐ Pain prevents me from walking more than 2 kilometres
- ☐ Pain prevents me from walking more than 1 kilometre
- ☐ Pain prevents me from walking more than 500 metres
- ☐ I can only walk using a stick or crutches
- ☐ I am in bed most of the time

Section 5: Sitting

- ☐ I can sit in any chair as long as I like
- ☐ I can only sit in my favourite chair as long as I like
- ☐ Pain prevents me sitting more than one hour
- ☐ Pain prevents me from sitting more than 30 minutes
- ☐ Pain prevents me from sitting more than 10 minutes
- ☐ Pain prevents me from sitting at all

Section 6: Standing

- ☐ I can stand as long as I want without extra pain
- ☐ I can stand as long as I want but it gives me extra pain
- ☐ Pain prevents me from standing for more than 1 hour
- ☐ Pain prevents me from standing for more than 30 minutes
- ☐ Pain prevents me from standing for more than 10 minutes
- ☐ Pain prevents me from standing at all

Section 7: Sleeping

- ☐ My sleep is never disturbed by pain
- ☐ My sleep is occasionally disturbed by pain
- ☐ Because of pain I have less than 6 hours sleep
- ☐ Because of pain I have less than 4 hours sleep
- ☐ Because of pain I have less than 2 hours sleep
- ☐ Pain prevents me from sleeping at all

Section 8: Sex Life (if applicable)

- ☐ My sex life is normal and causes no extra pain
- ☐ My sex life is normal but causes some extra pain
- ☐ My sex life is nearly normal but is very painful
- ☐ My sex life is severely restricted by pain
- ☐ My sex life is nearly absent because of pain
- ☐ Pain prevents any sex life at all

Section 9: Social Life

- ☐ My social life is normal and gives me no extra pain
- ☐ My social life is normal but increases the degree of pain
- ☐ Pain has no significant effect on my social life apart from limiting my more energetic interests e.g. sport
- ☐ Pain has restricted my social life and I do not go out as often
- ☐ Pain has restricted my social life to my home
- ☐ I have no social life because of pain

Section 10: Travelling

- ☐ I can travel anywhere without pain
- ☐ I can travel anywhere but it gives me extra pain
- ☐ Pain is bad but I manage journeys over two hours
- ☐ Pain restricts me to journeys of less than one hour
- ☐ Pain restricts me to short necessary journeys under 30 minutes
- ☐ Pain prevents me from travelling except to receive treatment

Score: / x 100 = %

Scoring: For each section the total possible score is 5: if the first statement is marked the section score = 0, if the last statement is marked it = 5. If all ten sections are completed the score is calculated as follows:

Example: $\frac{16 \text{ (total scored)}}{50 \text{ (total possible score)}} \times 100 = 32\%$

If one section is missed or not applicable the score is calculated: $\frac{16 \text{ (total scored)}}{45 \text{ (total possible score)}} \times 100 = 35.5\%$

Minimum Detectable Change (90% confidence): 10%points (Change of less than this may be attributable to error in the measurement)

Source: Fairbank JCT & Pynsent, PB (2000) The Oswestry Disability Index. *Spine*, 25(22):2940-2953.
Davidson M & Keating J (2001) A comparison of five low back disability questionnaires: reliability and responsiveness. *Physical Therapy* 2002;82:8-24.

*Note: Distances of 1mile, ½ mile and 100 yards have been replaced by metric distances in the Walking section.

A2. Reliability and validity of the viscoelastic model

Reliability

Five LBP and five healthy subjects were re-tested within a week from the first test to evaluate the reliability of the data provided by the experimental model. Pearson product moment coefficient was used to determine the reliability for the stiffness, damping and moment of inertia coefficients. Participants were asked to re-perform the 20° initial tilting trial. In the test-retest analysis, the stiffness and moment of inertia data were found to be highly reliable ($r > 0.9$) Reliability was low ($r < 0.2$) for the damping coefficient, but this was due to the value of this coefficient being almost zero. Mean and SD for the three coefficients in the test and re-test trials are shown in Table A2-1.

Table A2-1: Mean and SD for the stiffness, damping and moment of inertia coefficients measured in 5 healthy and 5 LBP subjects. The re-test trial was performed within a week after the first test

	HEALTHY			LBP		
	K	B	J	K	B	J
Test	28.76 ± 4.36	0.11 ± 0.06	5.11 ± 1.00	47.25 ± 4.73	0.63 ± 0.32	6.39 ± 1.45
Re-Test	29.15 ± 4.73	0.23 ± 0.21	5.12 ± 0.96	46.53 ± 7.24	0.60 ± 0.57	6.55 ± 1.49

Validity

Correlation between moment of inertia (j10 and j20) and body mass of subjects was evaluated using Pearson product moment coefficient in order to check validity of the modelling as proposed by Hodge and Van Horn (2009). Results showed significant correlation between the parameters for all subjects in both trials. (Correlation between J10 and body mass, $r = 0.69$; Correlation between J20 and body mass, $r = 0.73$). Furthermore correlation was also performed between j10 and j20 trials; this correlation was found significant ($r = 0.67$) confirming the validity of the model.

A3. Curvature of the spine analysis

Data from the curvature of the spine procedure were analysed in order to obtain the magnitude of the thoracic kyphosis angle, lumbar lordosis angle, and to plot the thoracolumbar curvature of the spine (Figure A3-1) as it was done by Singh et al. (2010) in their study.

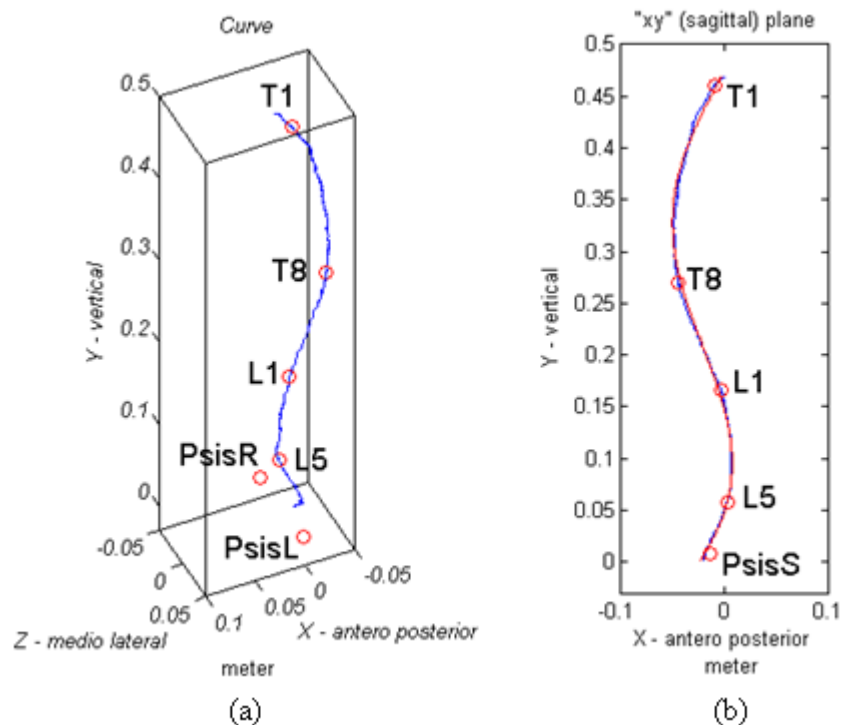


Figure A3-1: (a) 3-dimensional view of the thoracolumbar curvature; (b) Sagittal plane of the curvature

The digitised points were transferred from the coordinate system of the electromagnetic system source (global coordinate system) to a coordinate system fixed to the pelvis (local coordinate system). The local coordinate system was built with the points digitised in the left and right PSIS and T1 vertebrae. The origin was the midpoint between the two PSISs. The data of the thoracolumbar curvature were then fitted in a set of fifth-order polynomial equations. The thoracic kyphosis angle and the lumbar

lordosis angle were calculated using the digitised point. A polynomial equation was calculated with the data from the digitised points. Tangents at T1, L1 and L5 were derived by determining their derivatives. Thoracic kyphosis angle was defined as the angle between the intersection of the tangents derived for T1 and L1. Lumbar lordosis was defined as the angle between the intersection of the tangents derived for L1 and L5. The lordosis curvature was defined as positive and kyphosis curvature as negative. Matlab has been used for all the calculations. In Figure A3-1 an example of the thoracolumbar curvature for one participant in three dimensions is shown.

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